

BASIC MEDICAL RADIATION PHYSICS

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To James L. Weatherwax
teacher, scientist, friend.

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My family has been remarkably tolerant during a prolonged period; without their understanding and cooperation this book could not have been completed.

Preface

This book was developed from a course presented to radiologists and other physicians in the Philadelphia area for over four decades. The course was originated in 1924 by James L. Weatherwax, M.A., and has been taught by the author during the past 16 years. Available texts have been assigned, but it has become increasingly necessary to supply supplementary outlines to clarify physical concepts. These outlines have been enthusiastically received, encouraging the conclusion that a new basic radiation physics text is needed.

Two types of textbooks are possible, each serving its own useful purpose. The first definitively discusses physical principles of the main medical applications of ionizing radiation, including x-ray diagnosis, radiation therapy, dosimetry, diagnostic nuclear medicine, and so forth. Twenty-five years ago one could reasonably attempt this task in a single volume; in 1969, a group of rather large books would be required. Several other authors have recognized this fact, and are responding to the need for such specialized texts.

The present text is of the second type: a single book of reasonable size and cost, covering the basic concepts and tools of medical radiation physics. Such a text by itself can satisfy the primary needs of many readers, and in addition provide a useful conceptual bridge to more advanced presentations. The main intended audience includes physicians working in radiology and nuclear medicine, medical students, and other persons in the biologic sciences. In addition, engineers and other physical scientists may use the book to obtain a general idea of the properties, uses, and tools of ionizing radiation in medical and some industrial applications. This audience and the need for conciseness preclude rigorous and complete discussion of physical principles. Emphasis is therefore on conceptual clarity with scientific accuracy. Some important topics of radioactivity, x-ray attenuation, and dosimetry are treated in more detail, but with a minimum of mathematics.

Experience has shown a review of basic physics terms and concepts to be useful. Chapters 1 and 2, therefore, include a brief elementary summary of atomic structure, energy, radiation, and electricity. In addition, basic x-ray machine principles are described for the beginning user. These two chapters may be omitted by some readers, but can provide a concise and convenient review for those interested. Chapters 3 through 12 deal with x-rays and radioactivity; chapters 13 through 15, with radiation protection. The final chapter covers miscellaneous topics not readily included elsewhere.

Some final comments are of interest: Most figures are schematic rather than detailed, for clarity. References are kept to a minimum by citing source books and survey articles; however, some basic and historical articles are mentioned. A glossary of terms frequently encountered in medical radiation physics is included. An appendix lists sources of important information regarding dosimetry, radiation safety, and other important topics.

Philadelphia
January, 1969

Leonard Stanton

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**BASIC
MEDICAL
RADIATION
PHYSICS**

1

Matter, Energy, and Radiation

Certain basic principles are essential to an understanding of ionizing radiation and its applications, so we shall provide a brief review. The present chapter develops concepts relating to the nature and structure of matter, atomic and nuclear energy levels, and the wave and particle nature of electromagnetic waves. Chapter 2 then deals with basic electricity and the fundamental parts and operation of x-ray machines.

MATTER

Since before ancient Greek times men have wondered about the nature of matter. Is it continuous or porous? How can water dissolve some materials with apparent shrinkage in the total volume? Why does iron rust, while gold does not? These and many similar questions led first to pure speculation, later to systematic investigation, and ultimately to scientific development.

In this section we briefly review the basic nature and structure of matter.^{1,2}

General Nature of Matter

For our purposes everyday experience is more satisfying in defining matter than abstract reasoning. We use the term "matter" to refer to objects of tangible nature which resist changes in their motion (i.e., exhibit inertia). Thus, a steel ball and a feather are both matter. The steel ball is harder to stop abruptly than the feather, so we say it has more inertia and hence more matter in it. Material objects are acted upon by four basic kinds of forces: mechanical, electrical, magnetic, and gravitational. In general, they resist the abrupt intrusion of other objects, for example, a table in the dark or an automobile at an intersection.

Experience also tells us that matter comes in solid, liquid, and gaseous forms or states. In addition, exotic states exist under special

conditions, such as "plasma" which occurs in atom bomb explosions and the interior of stars.

MIXTURES AND SUBSTANCES. As they occur in nature, objects almost always contain two or more kinds of matter mixed together. This results from many causes. First, the earth's entire crust was produced by solidification of a liquid mixture of materials, a process repeated in volcanoes to the present day. Second, the action of wind and rain results in erosion of land, moving enormous quantities of mineral and organic matter to the plains and river beds, ultimately to the sea. Finally, living creatures construct superbly organized complex mixtures; after death they leave us such aggregates as the wood of plants, bone of animals, coral, marble, etc.

The components of mixtures range in size from very large to very small. If particle sizes are large, they may be plainly discernible. While ordinary dirt is an example of this, so are beautiful grained onyx and marble stones as well as some gems whose impurities lend great interest and beauty. With smaller particle sizes suspensions result, of which gels and sols are of biological interest. Finally, true solutions involve much smaller particles. Because water is such a powerful solvent, aqueous solutions are most familiar, but air is also a solution.

Mixtures may or may not be easily separable into their components. Perhaps the most basic properties of mixtures are their composition, continuously variable over a wide range, and their properties, reasonably similar to those of their constituents.

Much confusion reigned until chemists isolated pure materials, called *substances*. These materials have unique physical properties, such as melting and boiling points, as well as various electrical, optical, and other characteristics. No matter how tiny a sample one takes of a pure substance, he finds the same physical properties.

COMPOUNDS AND ELEMENTS. There are two kinds of substances: compounds and elements. Elements differ from compounds in that they can neither be broken down into two or more simpler substances by ordinary chemical procedures nor synthesized by combining other elements. Only about 103 elements are currently known. The number of possible compounds or chemical combinations of elements, however, is virtually unlimited.

Just as mixtures are combinations of substances, so compounds are evidently combinations of elements. Here, however, the resemblance abruptly ends. The early chemists quickly noted basic differences, of which the following three are most important:

1. Unlike those of a mixture, a compound's characteristics are in general completely different from those of its constituent

- elements. Thus, *liquid* water consists of *gaseous* hydrogen and oxygen; colorless crystalline sugar, of black carbon plus gaseous hydrogen and oxygen, etc.
2. Unlike a mixture, a compound *always* has the same composition by weight of its constituent elements (definite proportions law). If extra of any one constituent is added during synthesis, it is always left over.
 3. Sometimes the same elements combine to produce more than one compound. In such cases, for a given combined amount of one element in both compounds, the amounts of the other elements vary as ratios of whole numbers (multiple proportions law).

Example (1): Water and hydrogen peroxide: 2.016 g of hydrogen are combined with exactly 16 and 32 g of oxygen, respectively. ($32/16 = 2/1$.)

Example (2): Formaldehyde and methyl alcohol: 2.016 and 4.032 g of hydrogen are combined respectively with exactly 12 and 16 grams of carbon and oxygen.

MOLECULES AND ATOMS. Several conclusions follow inescapably from these observations. First, compounds must be very special combinations to yield such surprisingly new properties. This is emphasized by the vast number of diverse compounds, each with unique properties, producible from so few (103) elements. Second, the unique composition by weight of compounds indicates elements combine in *simple* units to produce *complex* units of compounds; these units are called atoms and molecules, respectively. The law of multiple proportions strongly reinforces this concept. These ideas were expressed by Dalton in the Atomic Theory proposed in 1805.

Table 1 shows the elements arranged in order of increasing "atomic number" (symbol Z). This term is the sequence in the table determined by the element's atomic mass A (taking that of oxygen as 16), making certain corrections to yield an arrangement in which chemically similar elements fall into the same vertical columns. Table 1 is a simplified version of the familiar periodic table of the elements.*

Measurements have shown atoms are held to each other in molecules by bonds of various strengths. Some bonds are stronger than

* We have retained the older convention of setting A for oxygen as 16.0000, making A of carbon 12.01115. A newer convention assigns 12.0000 to carbon, which is a useful procedure in basic physics work. The distinction is of no practical importance in clinical radiologic physics.

Periodic Table of the Elements

1 H 1.0080																	1 H 1.0080	2 He 4.003
3 Li 6.940	4 Be 9.013											5 B 10.82	6 C 12.011	7 N 14.008	8 O 16.0000	9 F 19.00	10 Ne 20.183	
11 Na 22.991	12 Mg 24.32											13 Al 26.98	14 Si 28.09	15 P 30.975	16 S 32.066	17 Cl 35.457	18 Ar 39.944	
19 K 39.100	20 Ca 40.08	21 Sc 44.96	22 Ti 47.90	23 V 50.95	24 Cr 52.01	25 Mn 54.94	26 Fe 55.85	27 Co 58.94	28 Ni 58.71	29 Cu 63.54	30 Zn 65.38	31 Ga 69.72	32 Ge 72.60	33 As 74.91	34 Se 78.96	35 Br 79.916	36 Kr 83.80	
37 Rb 85.48	38 Sr 87.63	39 Y 88.92	40 Zr 91.22	41 Nb 92.91	42 Mo 95.95	43 Tc 99	44 Ru 101.1	45 Rh 102.91	46 Pd 106.4	47 Ag 107.880	48 Cd 112.41	49 In 114.82	50 Sn 118.70	51 Sb 121.76	52 Te 127.61	53 I 126.91	54 Xe 131.30	
55 Cs 132.91	56 Ba 137.36	57-71 see La Series	72 Hf 178.50	73 Ta 180.95	74 W 183.86	75 Re 186.22	76 Os 190.2	77 Ir 192.2	78 Pt 195.09	79 Au 197.0	80 Hg 200.61	81 Tl 204.39	82 Pb 207.21	83 Bi 209.00	84 Po 210	85 At 210	86 Rn 222	
87 Fr 223	88 Ra 206.05	89-103 see Ac Series																

Lanthanide Series	57 La 138.92	58 Ce 140.13	59 Pr 140.92	60 Nd 144.27	61 Pm 145	62 Sm 150.35	63 Eu 152.0	64 Gd 157.26	65 Tb 158.93	66 Dy 162.51	67 Ho 164.94	68 Er 167.27	69 Tm 168.94	70 Yb 173.04	71 Lu 174.99
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Actinide Series	89 Ac 227	90 Th 232.05	91 Pa 231	92 U 238.07	93 Np 237	94 Pu 242	95 Am 243	96 Cm 247	97 Bk 249	98 Cf 251	99 Es 254	100 Fm 253	101 Md 256	102 No 254	103 Lw 257
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others, requiring more work to separate the atoms. In general, these amounts of work are small by radiologic standards, of the order of 1 to 20 electron volts (eV) (Table 6). Since x-rays can release *thousands* of eV, they can potentially break any chemical bonds. Visible and even ultraviolet light release only about 2 to 4 eV, so they can break only weaker chemical bonds.

Components of Atoms

In 1805 Dalton believed atoms were indivisible, electrically neutral particles. (The word atom itself, from Greek, means indivisible.) However, it has since become indisputable that atoms have component parts. Three subatomic particles have been found: electrons, protons, and neutrons. (In addition, a large number of others have at least transient existence. These are beyond the scope of this book.) We shall now briefly discuss their nature and discovery.

ELECTRICAL NATURE OF MATTER. Long ago men observed phenomena which strongly indicated matter had electrical properties. Lightning is essentially electrical current flowing between clouds, and occasionally between clouds and the earth. The passage of lightning through air was shown by Franklin to be an electrical current and conclusively showed even air could supply and transport electricity. The discovery of "electrification" by the ancient Greeks (Chap. 2) had already shown matter to be "electric" in nature. *Electrically charged objects* are defined here as those able to attract small bits of *any* isolated material. Electric currents are movements of such charged objects, called "charges" for short (see Chap. 2).

Chemical batteries generate electricity purely as a result of insertion of dissimilar metals into certain water solutions. This certainly requires that electrical charges be involved in chemical changes. Faraday passed electric currents through certain solutions and established that a definite amount of charge always liberates one gram equivalent weight of element from solution during electrolysis (Fig. 1). This strongly indicated that chemical bonds involve transfer of charges between atoms.

Around the turn of the century extensive work was carried out passing electric currents through gases at low pressure. Figure 2 illustrates the setup used by Thomson to study "canal" rays. These are positively charged particles of the gas employed in the experiment. These gas particles acquire charges just like atoms involved in electrolysis. Furthermore, the total charges acquired per mole by different gases bear simple integral ratios to each other. This indicates that some fundamental atomic charge is involved in electrical discharge through all gases.

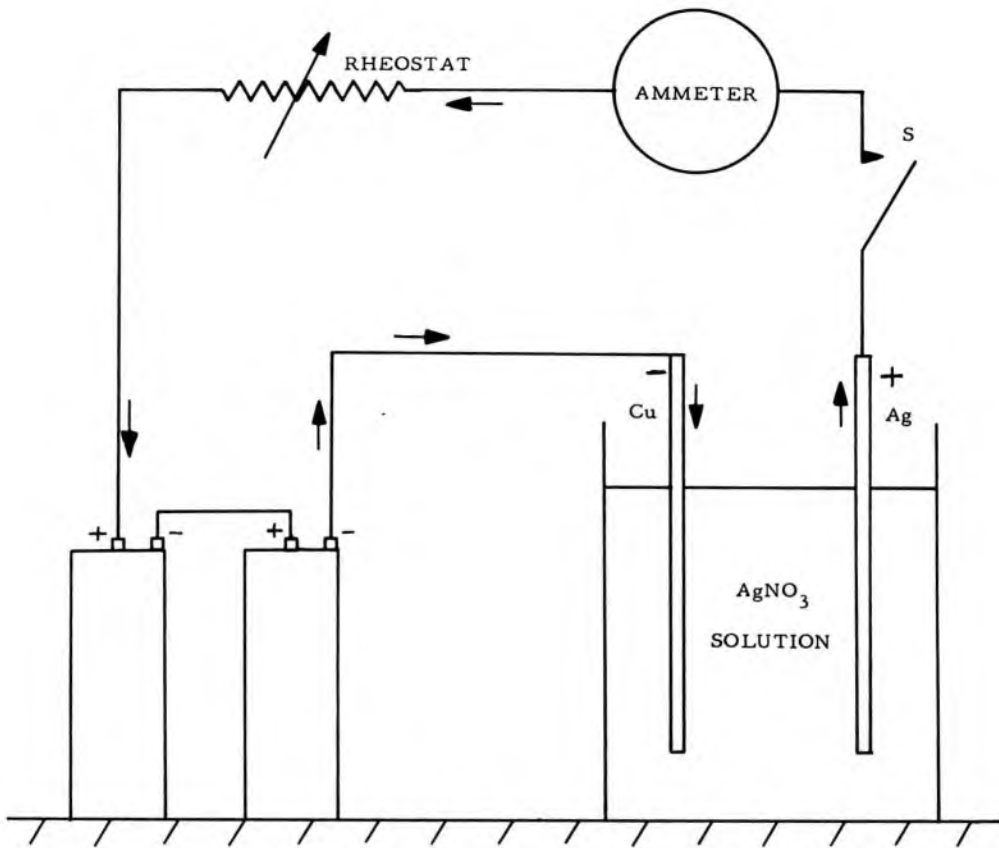


Fig. 1. Electrolysis suggests matter is electrical in nature. When switch S is closed, current flows as shown through the silver nitrate solution. Actually, plus charges travel from the silver to the copper electrode. After a time the copper electrode is coated with silver. The amount of silver gained by the copper is exactly the amount lost by the silver electrode—no silver is lost from the solution. Furthermore, the amount transferred is exactly proportional to the electric charge flow (current \times time). Careful study with many materials shows definite charges are associated with the transfer of each atom during electrolysis. Later work has shown this is either the electronic charge or simple multiples of it.

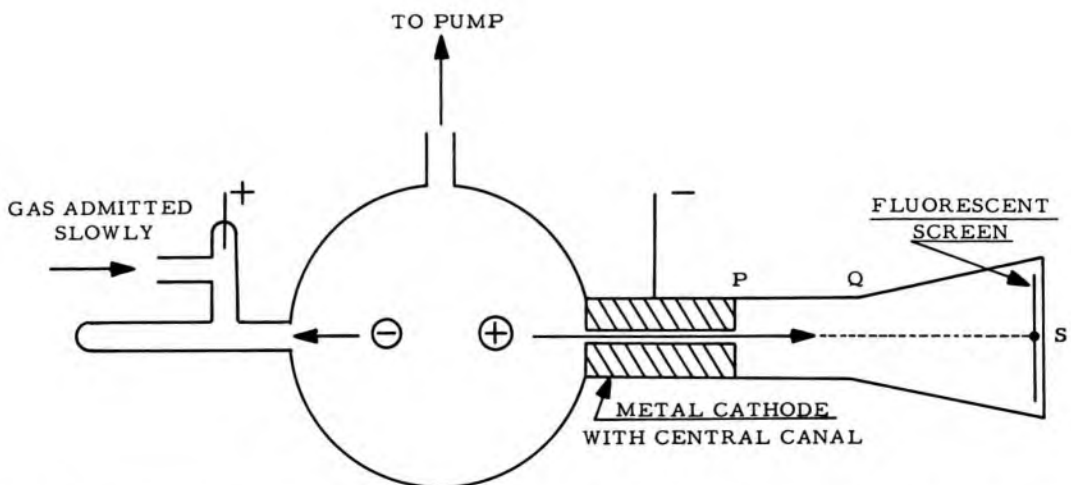


Fig. 2. Thomson's apparatus for studying positive gaseous ions, canal rays. The high voltage ionizes some gas molecules (p. 20). Positive ions

ELECTRONS. Even more surprising were results of experiments with negatively charged particles produced in gases at very low pressures. These "cathode rays" flow from the minus or negatively charged electrode (cathode) of highly evacuated glass tubes. As indicated in Figure 3 (top), these rays can produce fluorescence (visible light) upon striking objects, thereby revealing their point of impact. They have both mass and electric charge, as indicated by the experiments shown in Figure 3 (center and bottom). Their mass was found to be very low, about $1/1836$ as great as that of the lightest canal rays (from hydrogen), assuming the charge magnitudes to be the same. They are identical for *all* gases and cathode materials, a very surprising and significant result. These very light negative particles are called electrons.

Another related phenomenon is the production of "photoelectric current." This consists of minus charged particles ejected from some metal surfaces when they are irradiated by light (Fig. 4). These particles are found to be identical with the electrons of cathode rays. Since they must originate in the metal, this again proves metals contain electrons, which can be liberated by light. The work of Thomas A. Edison also showed they are liberated by heating metals as well. This effect is used in x-ray tubes to supply electrons (Chap. 2).

In 1909 Millikan measured the electron's charge directly and helped prove beyond a doubt that there is the same tiny electrically charged particle in *all* atoms. From this and much other work not mentioned, two conclusions become inescapable:

1. Atoms contain negatively charged particles of very tiny mass, called electrons. All electrons are identical in charge and mass, regardless of the materials from which they come. Electronic charge = 1.602×10^{-19} coulomb. See Chapter 2. Electronic mass = 9.11×10^{-31} kg.
2. Atoms are normally neutral but acquire negative or positive charges by either gaining extra electrons or losing them. When they acquire or lose charge, it can be only in integral multiples of the electron charge since electrons are indivisible.

Fig. 2 (cont.).

are attracted to the negative metal electrode on the right, containing a narrow hole in its center (called a "canal"), whereas negative ions go to the positive terminal. A few positive ions fortuitously pass through the canal and strike screen S, producing fluorescence and thereby identifying their path. These are canal rays. Application of electric and magnetic fields (not shown) in the tube between P and Q proves canal rays consist of positive ions of the gas. Each ion always has an integral number of plus charges of the same magnitude as the electron's charge.

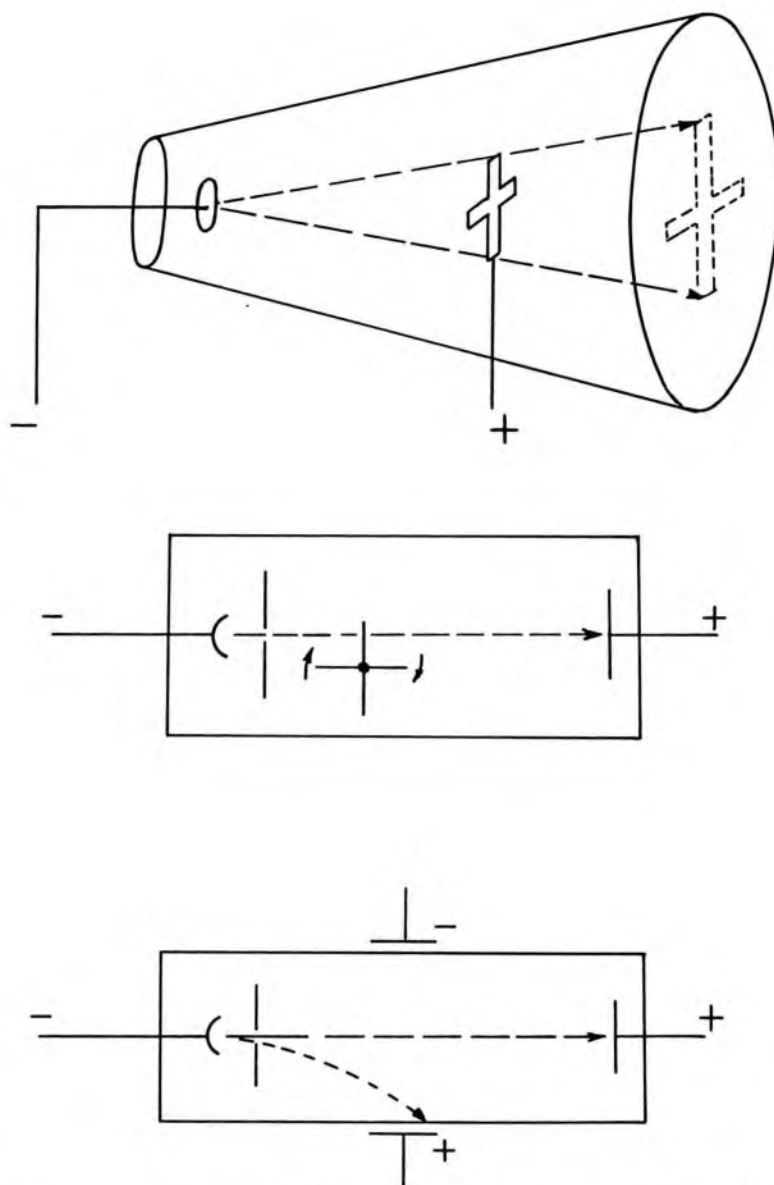


Fig. 3. Cathode ray particles. Top. Cathode ray particles have a minus charge, travel in straight lines, and produce visible fluorescence (light) upon striking glass and many other materials. Center. They have *mass*: the paddle turns when the voltage is applied and the particles strike the paddle. Bottom. They have a *minus* charge because of the way electric and magnetic fields deflect them (electric field shown). Their properties are *independent* of either the cathode material or the gas in the tube. This indicates they are particles released by *any* kind of atom when it is ionized.

The common hydrogen atom is neutral and contains only one electron. This is offset by a single positive charge identical in magnitude with that of the electron. The positive charge has been shown to be concentrated in a tiny particle called a *proton*. Its mass is about 1,836 times that of an electron by measurement.

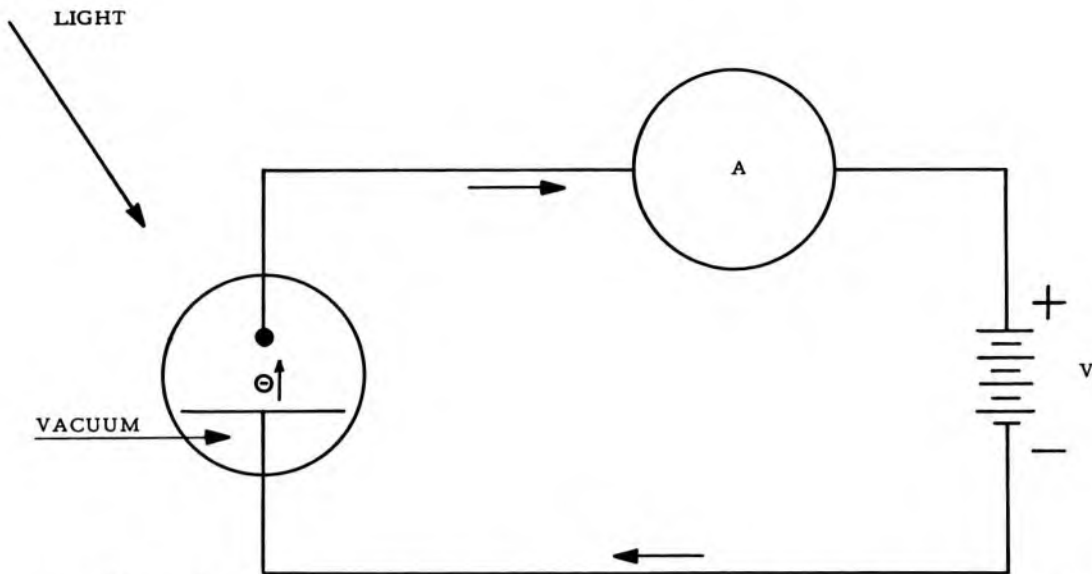


Fig. 4. Photoelectric cell operation. In darkness, no current is measured by A, a sensitive current meter. When light shines on the metal cathode surface, current flows as indicated by the arrows. Studies with electric and magnetic fields prove this current consists of particles identical to those of cathode rays. The voltage is employed to collect the photoelectric particles for measurement.

PROUT'S HYPOTHESIS. It is now evident that atoms normally contain electrons and protons in equal numbers, just as hydrogen atoms do. Recall that the atomic number of an element (symbol Z) is its sequence in the chemical periodic table of the elements. It is also the number of electrons or protons in one of its neutral atoms.

This fact leads to the idea that heavier atoms might simply be

TABLE 2. RELATIONSHIP BETWEEN ATOMIC WEIGHT AND ATOMIC NUMBER

Element	Atomic Number	Predicted Atomic Weight Protons & Electrons Only*	Actual	Ratio of Actual to Predicted
H	1	1.008	1.008	1.000
He	2	2.016	4.003	1.985
Li	3	3.048	6.940	2.28
Be	4	4.032	9.013	2.23
B	5	5.040	10.82	2.15
C	6	6.048	12.011	1.985
N	7	7.056	14.008	1.985
O	8	8.064	16.00	1.985
F	9	9.072	19.0	2.095
Ne	10	10.080	20.18	2.00
Cu	29	29.232	63.57	2.17
Sn	50	50.400	118.70	2.35
Pb	82	82.656	207.21	2.505

* Assuming only protons in nucleus.

hydrogen atoms packed together to yield more protons and electrons (Prout's hypothesis). However, this temptingly simple approach fails for two reasons:

1. If this were true, atoms would increase in atomic weight at a uniform rate, in steps of the weight of a hydrogen atom for each unit increase in Z . Thus, one might expect the sequence of element atomic weights to be as shown in the third column of Table 2. Actually, after hydrogen we see the actual atomic weight (column 4) is usually more than nearly *double* the predicted value.
2. As we shall see below, the protons are all packed into a tiny volume less than 1 million-millionth of a centimeter in diameter (10^{-12} cm). Without some nuclear bonding forces, the tendency for these mutually repelling charges to fly apart would be overwhelming.

NEUTRONS. By 1924 both Rutherford and Chadwick believed that there must be a neutral particle in the atomic nucleus. This could both account for the missing nuclear mass and, by hypothesized interactions with protons, act as a "cement" to hold the protons together. In 1932, Chadwick discovered the "neutron," and it has since been established as the other inhabitant of atomic nuclei. (Protons and neutrons are often referred to as *nucleons*.) We shall have more to say later about the properties and uses of neutrons in radiology.

Table 3 summarizes essential data regarding the three basic atomic particles.

Structure of Atoms

By the turn of the century, the existence and properties of electrons and protons were reasonably well-established. An important question then arose: what is an atom like? Its approximate diameter was estimated as an Ångstrom unit or so by 1910 (1 cm = 100,000,000 Ångstrom units. Symbol is Å; 1 cm = 10^8 Å). But was it a relatively large volume of fluid-like material in which the charges were embedded or mostly empty space separating tiny parts? This fundamental question was resolved by Rutherford in 1911. He and his group

TABLE 3. BASIC NUCLEAR PARTICLES

Particle	Atomic Location	Charge vs. Electron	Mass vs. that of Electron	Our Symbol
electron	orbits	-1	1	⊖
proton	nucleus	+1	1836	⊕
neutron	nucleus	None	1838	○

directed narrow beams of alpha rays (these are high speed helium atoms without their electrons) through thin mica or metal foils and observed the directions of emerging particles by noting where they produced light flashes on a fluorescent screen. Most of the particles went right through the foil, producing a central bright spot. Many, however, collided with the foil atoms and were deflected to the side. The resulting distribution of flashes over the screen was carefully analyzed. The pattern indicated that projectiles approached gold atom centers as closely as $1/10,000$ of an Å! This and other experiments indicated the following structure of atoms (Rutherford-Bohr theory).

1. The neutrons and protons are all located in a central tiny "nucleus," whose diameter is of the order of $1/10,000$ th that of the atom. This is an impressive demonstration that subatomic particles may exert surprisingly great forces. This effectiveness results from the fact that electrical forces are quite great at these short distances.
2. The electrons are excluded from the nucleus but held within the atoms, confined by the intense nuclear electrical attraction. The electrons are confined to certain locations around the nucleus, analogous to the planets' orbits about the sun.

The concept of electrons moving in circular orbits produced some theoretical problems and has since been abandoned. Classical electrical theory indicates an electron traveling in a circular orbit should radiate energy continuously and hence slow down steadily, and ultimately reach the nucleus. Actually, we now conceive of the electrons as a cloud of charges surrounding the nucleus, each in uniquely permitted energy levels (see p. 18).

THE NUCLEUS. The nuclear protons and neutrons are confined to an extremely tiny volume of incredibly great density. Because of their great proximity, one might expect protons to repel each other with enormous forces. It is believed, however, that other forces also exist between *all* nucleons, independent of charges—i.e., n-n, n-p, and p-p forces. These are believed to be small at distances of the order of Ångstrom units but increase to enormous values as one approaches nuclear particle separations. In general, a nucleus represents a balance between the finding forces between nucleons and the repulsive forces between protons which tend to make them fly apart. In some nuclei this balance is tenuous, and a good probability exists of nuclear disintegration. Such nuclei may be radioactive or otherwise unstable (Chapter 9). Others are more permanent; they are referred to as "stable" nuclei.

DEFINITIONS. The chemical nature of the atom is determined by the number of protons in its nucleus (the atomic number Z). This

discussed without reference to basic concepts of energy, work, force, and atomic energy levels. These concepts will now be considered.

Energy, Work, and Force

BASIC CONCEPTS. Energy is the ability to perform work. This concept is not unfamiliar in everyday life: one often says he lacks the energy to work or to perform a certain task. As used by the physicist, the term energy refers not only to such ability of a person, animal, or inanimate object but also to the "radiant energy" of x-rays and light, which are only indirectly associated with objects.

In a physical sense work involves the expenditure of energy to do one of two kinds of tasks:

1. *To overcome the opposition of objects to changes in their motion.* In essence, this means speeding up or slowing down objects whose inertia tends to keep them at rest or traveling at constant velocity. For example, we must work to push a stalled automobile on a level street. Once started, it tends to keep moving until brought to rest by work performed by brake or other friction forces (alternatively, a telephone pole or another automobile). Electrons acted on by electrical forces can be sped up or slowed down, just as an automobile.
2. *To overcome the opposition of objects to change in their association with others.* Objects all attract each other. The major forces are electrical, holding the atoms of solids together in relatively fixed arrays, and one must work to separate them. For example, one must work to cut metal, chop wood, carve a statue, or stretch a spring. For the same reason, overcoming friction involves the work of tearing or melting away tiny surface irregularities when adjacent surfaces move past each other.

A less potent but more familiar force is that of gravitation. The earth exerts a pull on all objects; we call this pull *weight*. To raise something we must work against this force; the greater the height, the more work to bring an object there. This fact comes as no surprise to anyone who has lived in a third-floor apartment without elevator service. More dramatically, it takes so much energy to overcome gravitation in sending a capsule into space that it costs roughly \$3,000 per pound. (Interplanetary migration is obviously not yet a practical solution to the problem of overpopulation.)

When one performs work on an object relatively free to move such as a baseball, golf ball, or electron, it acquires additional speed, and we say it receives energy of motion, or *kinetic energy*. Such kinetic

energy can be used wastefully, as in an automobile collision, or put to good use, as in a turbine electric generator driven by the rushing waters of Niagara Falls. On a molecular level, adding heat energy to an object does work on its molecules; they are made to move faster on the average, and we say the object is warmer.

Of most interest perhaps in radiology are the forces binding atomic particles. Forces between nucleons of a nucleus are enormous, so that cracking a nucleus takes considerable energy (millions of eV). (See Table 4 for summary of energy units.) On the other hand, electrons are bound by much smaller forces and hence may be separated relatively easily from their nuclei. Ordinary interactions between atoms, such as chemical, crystalline, and solution forces result from atomic interactions involving electrons in peripheral atomic locations. Such interactions involve only 1 to 20 eV of energy because the involved electrons are bound to their nuclei by relatively small forces. Other electrons which are closer to the nucleus are more strongly bound by electric forces and require more work (involving hundreds to thousands of eV) to remove them.

Energy is essential to all life processes, both in plants and in animals. Ultimately, our supply arises primarily from nuclear fusion which is the coalescing of hydrogen nuclei to produce helium nuclei. It requires temperatures of millions of degrees Kelvin, such as exist in the interior of stars as well as in the hydrogen bomb. Nuclear fusion occurs in the sun, which provides the earth with almost all its light and heat. This constant supply fortunately is stored continuously by plants in a form usable by animals as food. In addition, it has been stored as fossil fuels (coal, oil, and gas). Besides heating our buildings and running our vehicles, these fuels operate steam turbines to generate electricity, which in its great transportability and versatility is indispensable to modern civilization. It is generally agreed no nation can advance far technologically with limited electrical power resources.

UNITS OF ENERGY AND WORK. A variety of scientific units has been defined for measuring energy and the work it accomplishes (Table 4). Some useful units are the foot-pound and erg (centimeter-dyne) for mechanical energy, gram-calorie for heat, and electron-volt (eV) and joule (J) or coulomb-volt for electrical energy. All these units are applicable to any type of energy, since one form of energy is equivalent to any other and usually transformable to a considerable degree with the proper laboratory setup. The relationship of these units has been firmly established by both theory and experiment.

The top five units in Table 4 are of greatest interest in radiologic physics, the others in everyday work. The eV is simply related to the

TABLE 4. COMMON ENERGY UNITS, THEIR RELATIONSHIP TO ONE EV AND THEIR USE

Unit	Application
1 electron volt (eV)	The energy required to move an electron in vacuum from plus to minus across a difference of potential of one volt—order of energy needed to ionize outer orbit electrons.
1 kiloelectron volt (keV)	1,000 eV—order of energy needed to ionize <i>inner</i> orbit electrons. (10^3 eV)
1 megaelectron volt (MeV)	1,000,000,000 eV—nuclear disintegration levels; order of gamma ray energies. (10^6 eV)
1 billion electron volt (BeV)	1,000,000,000 eV—order of low cosmic ray energies. (10^9 eV)
1 erg = 628.1 billion eV	Unit of mechanical work. About the work done in lifting a 1 mg weight 1 cm at sea level.
1 joule = 10,000,000 ergs (10^7 ergs)	A useful, practical electrical work unit.
1 kilowatt-hour = 3,600,000 joules	The electric company's unit for charging customers for electric energy.
1 gram calorie = 4.185 joules	Unit for heat measurement. The heat energy needed to raise the temperature of 1 g of water from 4° to 5°C.
1 foot pound = 13,560,000 ergs	A more substantial unit of mechanical work than the erg; the energy expended in lifting a pound object one foot at sea level.

familiar chemist's unit of "kilocalorie per mole." One obtains eV by dividing kilocalories/mole by the factor 23.05. This was done to obtain the 1 to 20 eV figure for chemical binding energies, using standard chemical tables. The eV and its multiples are very useful in considering atomic and nuclear reactions. The erg is involved in defining absorbed dose in *rads* from ionizing radiation (Chapter 5).

Nuclear Energy Levels

As previously described, protons and neutrons are packed densely in nuclei. They are acted upon by tremendous forces because of their extreme closeness. We shall now briefly consider the forces and energy relationships involved.

Mutual repulsion forces between protons tend to make nuclei fly apart. Counteracting forces must therefore exist between nucleons to hold atoms together. All nuclei above those of simple hydrogen (${}^1\text{H}$) have neutrons, so these particles probably at least participate in binding action. Otherwise one should find other nuclides without neutrons, like ${}^2\text{He}$, ${}^3\text{Li}$, ${}^6\text{C}$, ${}^8\text{O}$, etc. *These have never been observed.* (See Table 2.)

It is generally believed that all nucleons (protons and neutrons) exert powerful attractive forces on each other at *close* distances; these forces fall off rapidly at separations approaching nuclear diameters (about $1/10,000$ Å). Internucleonic forces between protons (p-p forces) are evidently insufficient to overcome their electrical repulsion. Neutrons, however, are evidently much more adhesive, both to other neutrons (n-n forces) and protons (n-p forces). Neutrons thus serve the function of a "nuclear glue." However, their adhesive action is rather peculiar because nuclei with *too many* neutrons can become explosively unstable.

The nucleus normally tends to pack its protons and neutrons into certain preferred geometric arrangements. These involve a condition of maximum stability and minimum potential energy. It is analogous to a powerful steel spring, with its coils normally touching. If it is now stretched, it has potential energy and can do work (for example, run a grandfather clock for a week). Similarly, a nucleus can become distorted in shape also and thereby acquire extra potential energy. Of course, this distortion must be done by a nuclear reaction rather than a mechanical stretching. The nucleus can give up this energy *simply* by emitting a gamma ray upon the return of the nucleus to its normal undistorted condition; or, in some nuclides, it may actually disintegrate. (Gamma rays are photons like x-rays, usually of high energy. Photons are discussed in the next section.)

Why do we speak of "nuclear energy levels?" A coiled spring can have any energy from zero up to a maximum, depending on how much we stretch it. Nuclei however are different. As in so many aspects of atomic physics, they are "quantized." Essentially, this means the nuclear distortion can occur only in *steps*, unlike the stretching of a spring. These "energy levels" are uniquely characteristic of each particular nuclide. When the disturbed or excited nucleus gives up its energy, it can do so only in discrete steps corresponding to the characteristic energy levels. As a result, gamma rays emitted by the nucleus are also characteristic of the nuclide involved.

Excited nuclei are generally produced during nuclear upheavals such as fission and radioactive transmutations. The product nuclei are often left in an excited state and may emit gamma rays when they settle down to normal. (See Chapter 9.)

Electron Energy Levels

As previously indicated, an atom's electrons are in a cloud surrounding a tiny nucleus, and they cannot be considered to be rotating in literal planetary fashion. Where, then, are they located, and how can this explain atomic behavior? We shall now consider these questions.

BASIC ELECTRON LOCATIONS. Just as nuclei can be in ground and excited states, so can the electrons. Let us first consider an atom in the ground state, with its electrons in their "proper" locations.

There are certain *principal* locations, or energy levels. These are interchangeably called K, L, M, N, O, P, etc. and 1, 2, 3, 4, 5, 6, etc. They can be conveniently visualized as being at different distances from the nucleus, K or 1 being closest, the rest increasingly remote. The term "orbit," even though not literally correct, is retained because it is short and established by long usage.

In hydrogen and helium atoms (Figs. 5 and 6), the electrons are all in the K-orbit, and no others exist. However, as we increase Z from 3 through 103, adding protons to the nucleus, we must add electrons also. These fill the orbits from the inner orbits outward. The capacity for electrons is 2, 8, 18, 32, etc. for the K, L, M, N, etc. orbits, respectively.*

TABLE 5. ORBITAL STRUCTURE OF SEVERAL ELEMENTS OF BIOLOGICAL INTEREST

Element	Z	Number of Electrons in Given Orbit and Suborbit									
		K	L		M			N			
		<u>s</u>	<u>s</u>	<u>p</u>	<u>s</u>	<u>p</u>	<u>d</u>	<u>s</u>	<u>p</u>	<u>d</u>	<u>f</u>
Hydrogen	1	1									
Helium	2	2									
Carbon	6	2	2	2							
Nitrogen	7	2	2	3							
Oxygen	8	2	2	4							
Sodium	11	2	2	6	1						
Phosphorus	15	2	2	6	2	3					
Chlorine	17	2	2	6	2	5					
Potassium	19	2	2	6	2	6	1				
Calcium	20	2	2	6	2	6	2				
Iron	26	2	2	6	2	6	6	2			
Maximum number of electrons allowed in each suborbit		2	2	6	2	6	10	2	6	10	14

* The atomic number implies the total number of orbits. Thus, Z must be 2, 10, 28, and 60 to fill the K, L, M, and N orbits, respectively.

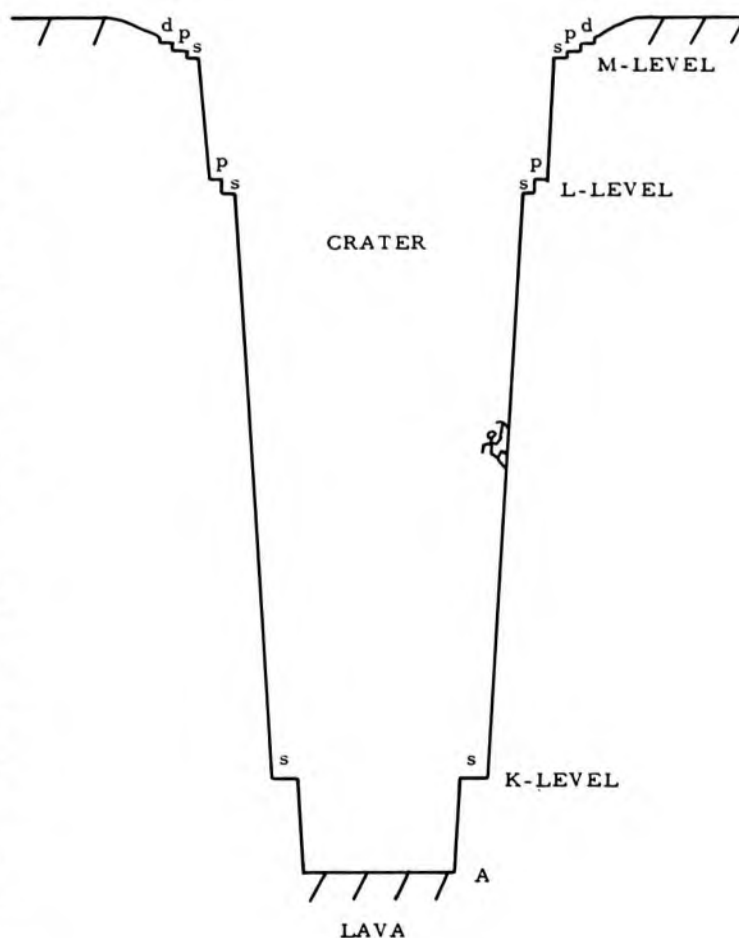
In addition to the principal locations, secondary locations (sub-orbits) exist. In K, L, M, N, and O orbits, their corresponding sub-orbits are labeled s; s,p; s,p,d; s,p,d,f; and s,p,d,f,g. These sub-orbits also have their allowed numbers of electrons, according to selection rules we cannot consider here. However, Table 5 shows the orbit and suborbit locations of electrons in several atoms of biological interest. The bottom row indicates the maximum allowed number of electrons in each suborbit.

Very rigid rules apply to electron locations. Strict "no parking" rules exist for all locations except the orbits, and even there the number of places is limited. No exceptions have been found, and this principle is accepted by physicists as one of the basic premises of modern science.

An analogy may help in comprehending this description. Consider Figure 7, showing a daredevil attempting to scale the nearly vertical

Fig. 7. Electron energy level analogy: climber in volcanic crater. Electrons normally cannot survive within the nucleus; analogously, the daredevil must stay out of the lava at level A. Three levels are shown where he can stop to rest: K, L, and M, corresponding to permissible electron levels. To climb out of the crater from the K level takes maximum work; from L, *less*; and from M, *least*. Hence, these are energy levels. (Note there are also sub-levels s, p and s, p and d.)

Since the climber must work to reach L and M locations from K, these are called higher energy levels. The highest possible level of the system is outside the crater altogether. Of course, the daredevil has an advantage over the electron: he can climb out under his own power whereas the electron must remain in one level unless assisted upward.



wall of an active volcano. In the side of this wall are tiny ledges at different levels, in up to seven groups (K through Q), of which only three are shown for simplicity. The first group has only one ledge, the second two, and the third three.

The man cannot survive at level A, nor can an electron in the nucleus because it would interact with one of the protons. The lowest place he can remain, therefore, is on the K ledge. This is the low energy location also, as the man must do maximum work to escape the volcano from this level. He now climbs upward but cannot rest until he reaches the L levels—Ls and Lp, where he may pause without danger of sliding back. Here he is at a higher energy level (more nearly out of the hole). Farther up, there are three levels at Ms, P, and d, and near the very top are final unlabeled and less sharply demarcated locations still within the crater; beyond this lies safety.

Atom orbits are analogous to the crater ledges. Atoms of simple elements correspond to shallow craters with only a few energy levels; those of more complex elements, to deeper ones with more energy levels. Why do we say electrons near the edge of the atom have more energy than those nearer the nucleus? It takes less energy to remove them from the atom. They are already remote from the attracting nucleus, and compared with more centrally located electrons, they have added energy.

It may be protested that this is a rather negative way of figuring energy. It is quite similar, however, to saying a man is more financially solvent because he owes less money, an analogy readily understood in these days of easy credit.

We may therefore summarize this discussion of basic electron location as follows:

1. Electrons occupy discrete locations in the atom, called energy levels.
2. These involve principal orbits, called K, L, M, etc., with sub-orbits s, s and p, s, p, and d, etc., respectively.
3. Electrons are forbidden from any other locations in the atom except in transit. These "no parking" rules are very rigidly maintained.
4. Normally the lower energy orbits and suborbits are filled before the outer orbits, as one considers atoms of increasing atomic number.

IONIZATION AND EXCITATION. Thus far we have discussed electron locations for atoms in their ground state or "non-energized" condition. However, even atoms cannot remain at peace without intrusion. The mere presence of other atoms in space makes inevitable collisions and bombardment by atoms, parts of atoms, and the resulting electro-

magnetic waves produced by such interactions (see below). Such collisions give the atom energy. This can make it move bodily as a result of a general interaction, or the intruder can interact specifically with the nucleus or one of the electrons. Nuclear interactions are discussed later in Chapter 9. We shall at this time consider only interactions with orbital electrons.

When an atom's electrons receive energy, one of two events can result: ionization or excitation. If an involved electron acquires enough energy to escape from the atom completely, the event is *ionization*. Alternatively, if the electron receives enough energy to move to a location of higher energy *within the atom*, the event is *excitation*.

Figure 8 illustrates these effects in an oxygen atom struck by a high energy electron. For simplicity the atom is shown in two dimensions. Also, all six L-orbit electrons are shown in the same level. (The 13.6 eV figure applies to the four p suborbit electrons.) Experiments have shown that 13.6 eV of work are required to completely remove an Lp electron, 532 eV to remove a K-orbit electron. Suppose the entering electron has 1,532 eV energy of motion. In the interaction (A), the K electron can be removed completely from the atom and be given velocity corresponding to 1,000 eV besides. The atom has then lost an electron completely and is *ionized*. When ionization occurs in this specific radiologic sense, we end up with a *plus*-charged atom and an escaped *negative* electron. The entering electron that started the entire upheaval almost always simply gives up some of its energy of motion and moves on and is essentially only an energy delivery agent. At the worst, it loses its kinetic energy and becomes a relatively slow "free electron" outside the atom with which it interacted.

An interesting question is this: suppose the K electron is struck by an incoming electron having only 520 eV? In this case, the energy available is insufficient to completely overcome the attraction of the nucleus for the K-orbit electron. One might expect the K electron to be raised to a level of $532 - 520 = 12$ eV, slightly beyond the L orbit of 13.6 eV. This, however, is impossible since only certain discrete levels are permitted—namely, 740 and 13.6, and 12 eV is different from these.

The result is that the K electron absorbs only $532 - 13.6 = 518.4$ eV and rises to the permitted 13.6 eV or L orbit (C). What of the extra 1.6 eV the entering electron had? It keeps this and emerges with it. The atom is left with its original electrons, but one is in a higher energy level. The atom has acquired 518.4 eV of energy and is called "excited." More generally, outer orbits have not only the standard suborbits but other allowed positions as well, so atoms and molecules can become excited by exposure to other atoms, light, and heat. This is the basis for many chemical effects of biological importance.

To summarize: atoms can be ionized or excited when intruding particles interact with orbital electrons. Ionization involves removing one of these electrons completely, leaving the atom deficient in an electron and hence positively charged. (The released electron and its parent atom are quite logically called an "ion-pair.") Excitation of the atom results when too little energy is available from the interaction to overcome the attractive forces binding the electron to the atom. However, the electron is given enough energy to jump to a permitted higher level, located either within the same orbit or one more remote from the nucleus. In both cases the atom has received extra energy which it can later release as heat, electromagnetic waves (light, x-rays, etc.), or chemical energy.

CONSEQUENCES. Up to this point we have considered only the ionization and excitation events themselves. It is natural now to investigate what happens subsequently, not only to the atom involved but to others in the neighborhood as well.

The ionized or excited atom can normally remain in these states for only a relatively short time, of the order of a microsecond (millionth of a second).^{*} In an excited atom the gap left in the old orbit is quickly replaced by either the same or another electron. When this happens, there is complete restitution of energy. Thus, in Figure 8 the (C) electron or another in the L orbit returns to the K orbit. The 518.4 eV of energy originally acquired in the excitation event is now released as a *single photon* of this energy. (The nature of photons is discussed in the next section.)

In an ionized atom, the lost electron may be some distance away from the original atom. (It can be many millimeters away in super-voltage x-ray therapy.) However, there are always free electrons, essentially unattached to any atom, in all materials. (These are, of course, associated with equal numbers of positive ions and generally arise as a result of natural radioactivity and cosmic rays.) Acquisition of such a free electron can return the ionized atom to neutrality. For example, in Figure 8 the electron (A) can be replaced by the new electron's going directly to the K orbit.

In this case, a single photon of all the originally acquired 532 eV results. (The meaning of photons is discussed on page 30.) A moment's reflection, however, shows that an alternative procedure could also occur, in *two* steps. First, an electron could move from the L to the K orbit within the atom. Then the outside electron could replace

^{*} Certain crystals are exceptional and exhibit longer delays and even electron trapping phenomena. These generally involve multiple, rather than single atoms, however.

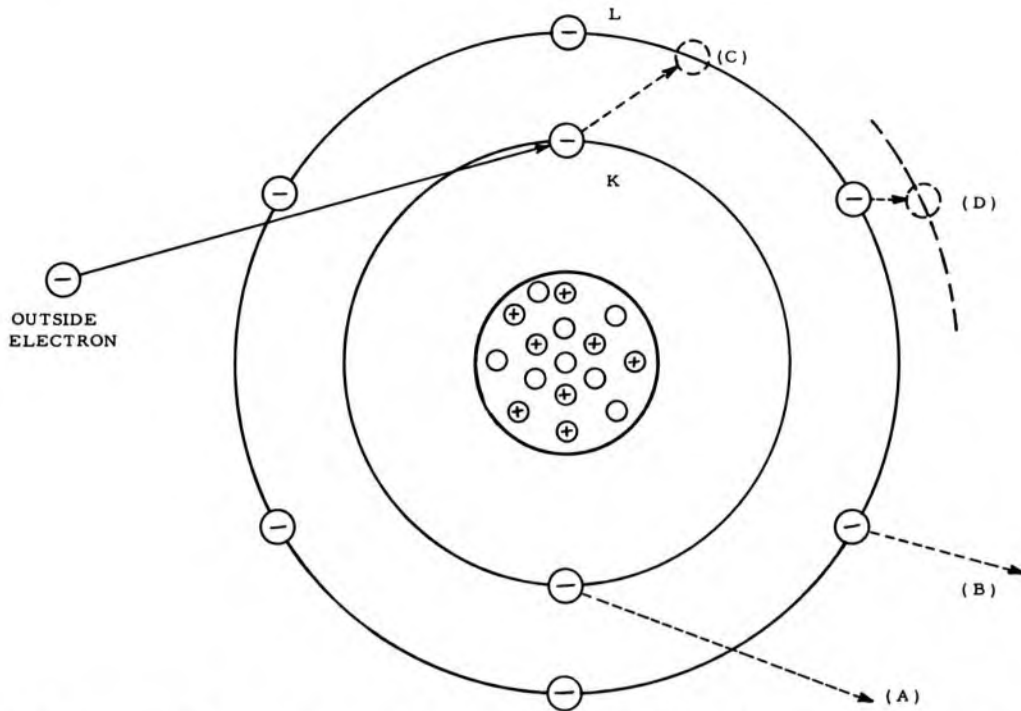


Fig. 8. Collision of a fast external electron with electrons of an oxygen atom. Ionization: it takes 532 eV to *completely remove one K electron*, (A); it takes only 13.6 eV to *completely remove one L electron*, (B). Excitation: it takes $(532 - 13.6)$ or 518.4 eV to move a K electron to the L orbit, (C); (D) there are also other permissible levels above the L orbit (dashed line). These are involved in producing visible light following excitation by heat and electrical conduction in gases. In both processes, characteristic radiation results upon restoration of the atom's original condition.

the resulting vacancy left in the L orbit. The result is the same, and both are possible events. Note that in the second restoration procedure two photons rather than one are emitted: 518.4 and 13.6 eV, one for each of the electron transfers; the total energy, however, is the same: 532 eV.*

This description tells only part of the story though, since it is limited to the atom itself. In general, the atom is both associated with and surrounded by other atoms. In its excited and ionized state, the atom's shape and bonding forces are altered; consequently, its chemical behavior may be correspondingly altered, affecting both chemical and crystalline lattice bonds. In addition to these local effects from the atom's acquisition of energy, more remote atoms may be ionized

* The "Auger effect" (Chap. 4) competes with characteristic photon emission when ionized atoms are restored. Outer orbit electrons may then be released by the atom instead of photons. The Auger effect is most important in low Z materials, less significant in high Z materials.

and excited by the electron ejected from the original atom during ionization. We shall have more to say about both of these effects when we discuss x-ray attenuation.

RADIATION

We use the term radiation to mean energy emanating from a point. The word is derived from the Latin *radium*—spokes of a wheel leading outward from the hub in the center. Energy can radiate from a point by two basic means: as energy of motion of traveling material particles (kinetic energy) and as waves.

An example of radiation of material particles is shrapnel from an exploding antipersonnel shell. The shell fragments travel outward in all directions with bullet-like speed; if the shell explodes in air these fragments convey lethal energy to life within its range. Material particle (“corpuscular”) radiation is also important in radiology. It usually consists of subatomic particles released in nuclear reactions. Corpuscular radiation includes neutrons as well as alpha and beta rays released from radioactive nuclides.

In this section we discuss waves primarily; corpuscular radiation is covered in Chapter 9. We shall first describe wave properties generally and then consider electromagnetic waves in particular.

Waves

When objects are struck or made to vibrate, disturbances generally emanate from them; these are called “waves.” Such waves may convey energy to areas remote from the original location by means of intervening material. The initiating phenomenon may be a violent shock such as an explosion or earthquake. A single major impulse, called a “shock wave,” then travels outward. Negligible matter, however, is transferred; only the impulse is communicated. These shock waves can occur in solids, liquids, and gases. Examples are tidal waves accompanying earthquakes in land and water and the blast from an atomic bomb in air. Such single pulse waves transfer considerable energy and on occasion produce great damage and casualties.

Of greater interest here are *repetitive* waves, which vibrate an object many times rather than simply striking it once. Sound waves are perhaps the most familiar example of such oscillatory “matter waves,” as we designate them here. All such waves have three things in common: they arise in vibrations of gross particles, are communicated only by means of material media, and cause other material particles to vibrate correspondingly.

BASIC WAVE CONCEPTS. The basic concepts of wave generation and transmission are indicated in Figure 9. Consider a lucky boy with nothing to do but sit by a large pond and slowly vibrate a board up and down in the water about twice a second. To his delight, as the board rises and falls, he notes that the water also rises and falls, and a wave travels outward in all directions. Surprisingly, a cork at Q some distance away also rises and falls two times a second when the wave reaches it. It is not moved significantly in the direction of the wave travel but is simply made to move up and down by the wave's passage. When the wave crests A and A' arrive, the cork is at its highest; when the low portions (troughs) arrive, the cork falls to its lowest level. Thus, it is the wave's *passage* which has vibrated the cork.

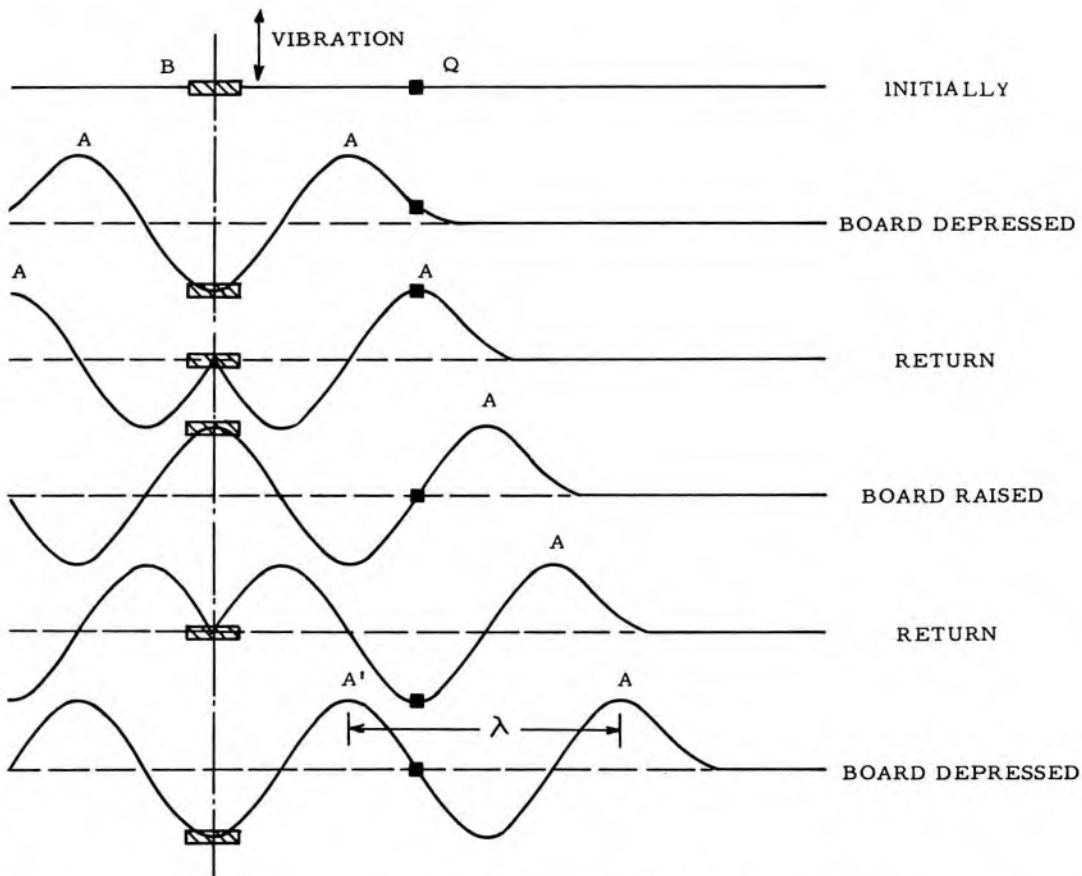


Fig. 9. Generation of a water wave. Slow up and down motion of board B forces the water to move up and down correspondingly. The crest A moves horizontally as the board vibrates, and a ripple spreads outward. The outward speed of the crest depends primarily on the material's elasticity and density. By the time the board is back to its depressed position however, a new crest A' has replaced A. The distance AA' is the crest separation and is designated λ , the wavelength. (Since the ripple must by its nature move λ in a cycle, this leads to the basic formula: $c = \lambda f$, c being the crest or wave speed and f , the rate of vibration.) Note that the cork at Q bobs up and down in a manner similar to the board's motion, but in general it reaches the top position at a different time.

The term "frequency" refers not only to the source vibration rate but also to both the number of crests reaching Q in a second and the resulting cork vibration rate. All are the same because the source caused the wave which in turn caused the cork vibration. This relationship is true of *all* wave motion, regardless of type.

The distance or length between successive crests helps describe the wave; it is therefore called the wavelength (λ). The speed of the wave (c) depends primarily on the density and elasticity of the vibrating material. It describes how fast the crests leave the source and is very simply related to the wave frequency (f) and wavelength (λ). The derivation of this relationship is simple. The crest at A' will reach location A in one cycle or complete event. This takes $1/f$ seconds (i.e., three cycles in one second means it takes $1/3$ second for each cycle). During this time the wave moves one wavelength or λ . Since velocity involves distance traversed per unit time, one may write:

$$c = \frac{\text{distance}}{\text{time}} = \frac{\lambda}{1/f} = f\lambda \quad (1-1)$$

Note that for a given wave f is constant. If the wave traverses a new medium so that the speed changes, λ changes correspondingly, but not f . The above simple relationships apply not only to all matter waves but to electromagnetic waves as well.

Matter waves of 20 to 20,000 cycles per second (one cycle per second is called a *hertz* (Hz); 1 million Hz = 1 MHz, a *megahertz*) may be detected by young people. They are called "sound." Ultrasound includes frequencies 20,000 Hz and greater. Frequencies of 1,000,000 to 15,000,000 Hz (one to 15 MHz) are used for ultrasound diagnosis (Chapter 17), with somewhat lower frequencies usually employed in therapy and industry. Very low frequency matter vibrations are *felt* rather than heard. They are nevertheless mechanically effective, and intense matter waves are capable of producing wear and tear on both inanimate objects and living organisms.

ELECTROMAGNETIC VERSUS MATTER WAVES. We have thus far discussed matter waves only. These not only require matter for their propagation but are themselves only disturbances in matter. Their speed ranges from about $1/5$ to 5 miles per second. While this is reasonably fast, a fast jet can readily exceed the speed of sound in air.

Electromagnetic waves include radio, light, x- and gamma-rays (Table 6). They differ from matter waves in several important ways, including the following:

1. *Speed*: 186,300 miles a second, about 100,000 times faster than matter waves, and theoretically faster than any material particle can travel, according to modern physical theory.

TABLE 6. ELECTROMAGNETIC WAVE RADIATION

Wave Category	Specific Kind	Typical Wavelength	Approximate Corresponding Photon Energy
Radio	Broadcast (1,000 kHz)*	30,000 cm	.000 000 004 eV
	T.V. (100 MHz)	300 cm	.000 000 4 eV
	Radar (3,000 MHz)	10 cm	.000 012 eV
Light	Infrared	12,000 Ångstroms (Å)	1 eV
	Visible	6,000 Ångstroms	2 eV
	Ultraviolet	3,000 Ångstroms	4 eV
X-Rays and Gamma Rays	Diagnostic	0.4 Ångstroms	30 keV [†]
	Orthovoltage Therapy	0.1 Ångstroms	120 keV
	Cobalt-60	0.01 Ångstroms	1,200 keV
	Linear Accelerator	0.06 Ångstroms	2,000 keV
	Betatron	0.0015 Ångstroms	8,000 keV
Cosmic Ray-Produced Waves	Photons produced by cosmic particles striking atoms on earth.	0.000 012 Ångstroms	1,000,000 keV and up.

* One cycle per second = one hertz (Hz)

1000 Hz = 1 kHz

1,000,000 Hz = 1 MHz

[†] These energies are NOT the maximum values present in the beams, which have a broad range of energies present. (See Chapter 3.)

2. *Medium*: unlike matter waves which by their very nature cannot traverse a vacuum, electromagnetic waves travel *best* (fastest and with least attenuation) in empty space with no atoms to get in the way.
3. *Origin and interactions*: matter waves originate in the mechanical interactions of usually electrically neutral objects, such as vocal cords and loudspeakers. Electromagnetic waves generally originate from interactions of charged parts of atoms and groups of atoms. (Gamma rays originate in nuclei which contain protons and are hence charged.) Furthermore, they can in turn interact with charges inside atoms and

groups of atoms. Radio waves vibrate the free electrons of antennae wires; light ionizes and excites *outer* orbital electrons of atoms; and x-rays interact with *all* orbital electrons of atoms, and even nuclei if their energy is sufficiently great.

As the name indicates, electromagnetic waves are both electrical and magnetic in nature. They are propagated as fluctuating electric and magnetic fields which reach out through space from a local electrical disturbance. For example, rapidly oscillating electrons in television transmitting antennae cause radio waves to radiate from them. These radio waves are received and used by television sets to display the edifying messages originating at the transmitting station.

Some useful radio waves have frequencies less than 1 MHz. Light wave frequencies are of the order of *500 million* times greater whereas those of diagnostic x-rays are about *6 million-million* times (6×10^{12} times) greater.

Electromagnetic Waves—Wave Nature

A few hundred years ago there was a lively dispute concerning whether light consisted of waves or particles, and no less an authority than Newton held they were discrete particles. It is now believed they exhibit *both* wave and particle properties. We shall show below the nature of the wave properties and in the remaining sections consider their particle properties and draw some practical conclusions.

DIFFRACTION AND INTERFERENCE. Whereas some waves can exhibit specific properties like polarization, *all* waves exhibit diffraction and interference.

Diffraction is the bending of waves around the edge of obstacles. To illustrate, in Figure 10 (top) a man has sought relief from a noisy jackhammer at (B) by moving to a position (S) behind the corner (C) of a substantial building. Even in an area without reflecting surfaces he finds some of the noise (A) bends around the corner, and that he must leave the area if he is to find a quiet location.

In another example, Figure 10 (bottom), light passes through a small round hole in an opaque card. One would, of course, expect a bright, round central spot on a screen placed at B, and this is indeed found. However, if the light consists of only one color, the spot is surrounded by diffraction rings, a set of concentric rings resembling a bull's-eye. With white light a series of rings of rainbows is obtained. This, too, is caused by diffraction. The alternating light and dark rings are respectively in locations of constructive and destructive interference, resulting from the bending of light around the edge of the hole. In the bright areas, "crests" of light waves from opposite edges of the hole arrive at the same time and add in brightness (constructive inter-

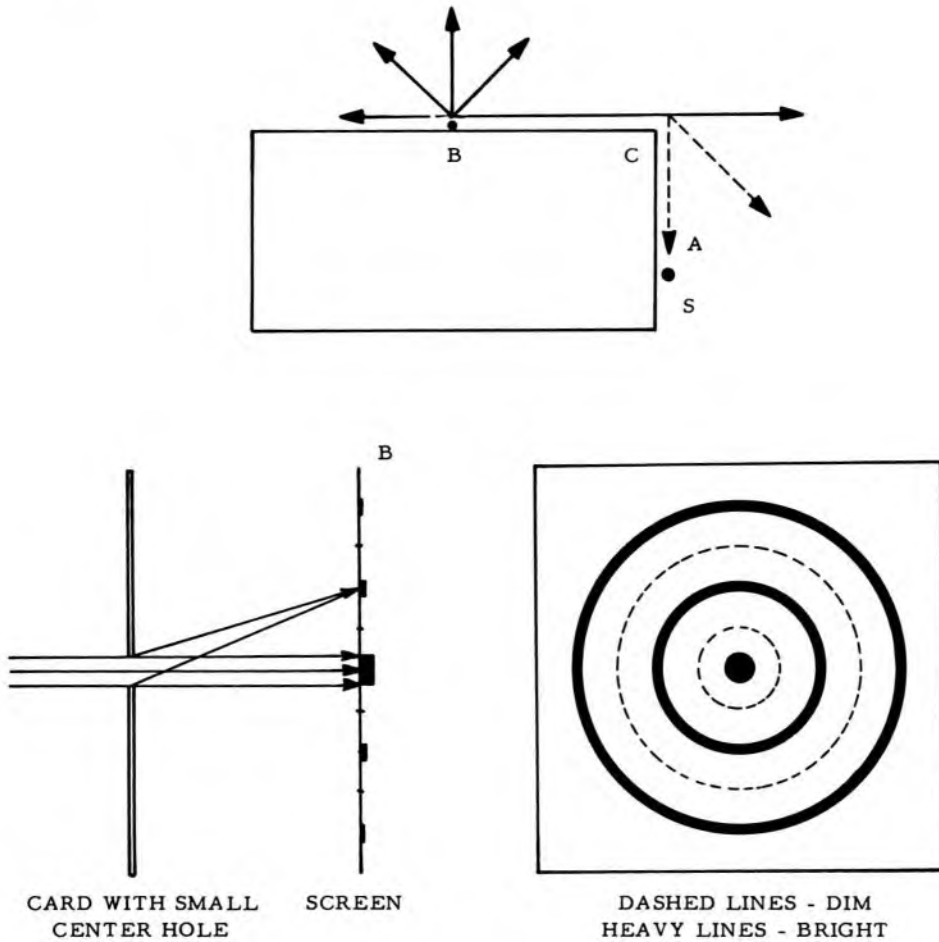


Fig. 10. Diffraction and interference. Top. Sound diffraction. The sound cannot penetrate the building significantly and should travel only in the directions shown by the solid lines. However, it bends around the corner C (dashed lines) and the man hears it. Bottom. Light diffraction and interference. The direct rays produce a bright center spot on the screen, as expected. However, light bends around the edges of the small round hole, so the deflected rays illuminate the screen beyond the central spot. Rays from opposite sides of the hole travel slightly different distances, so their crests arrive at different times. Sometimes the crests of the first ray arrive coincidentally with the depressions of the second; they then tend to cancel. Farther out from the central spot, the crests arrive together and produce a brighter image. We actually observe alternate bright and dim circular areas (bottom right) of constructive (solid lines) and destructive (dashed lines) interference.

ference); in between, crests interact with troughs, yielding dim areas (destructive interference).

X-RAY DIFFRACTION. Similar diffraction-interference effects are demonstrable with x-rays, but they are experimentally more difficult to produce because wavelengths are so short. In Figure 10 (bottom) the light rings are relatively dim compared with the direct beam and are hard to see if the hole diameter is too large (being the dim image

at the edge of a large bright field). In general, for diffraction to be manifest, the wavelength must not be too much less than the separations of the corners around which the beam is to bend. Medical x-ray wavelengths range from 0.5Å to less than 0.1Å. To do useful x-ray diffraction measurements, one must therefore employ corner separations of the order of 1 to 10Å. Fortunately, nature has provided regular separations of about 2Å in crystals, and these are used for x-ray diffraction measurements. One must therefore employ corner separation; x-ray diffraction is extremely useful in study of crystal and macromolecular structure. Experiments definitely prove that x-rays undergo diffraction and interference and hence act as waves.

X-ray diffraction is not important in clinical radiology, in which both diagnostic and therapeutic applications depend primarily on the *particle* characteristics of x-rays described below.

Electromagnetic Waves—Particle Nature

Unfortunately for those seeking simple answers, electromagnetic waves also interact with matter as particles do. However, they do not consist of the ordinary kind of particles, with mass, but rather of discrete bundles of energy, without mass. Particles of light or x-rays are called *photons*. The amount of energy in a photon is surprisingly related to the wavelength by the simple expression:

$$E = 12.4/\lambda \quad (1-2)$$

where E is photon energy in *keV*
 λ is wavelength in *Angstroms*

Actually $E = hf$, where h is the famous "Planck constant" (6.61×10^{-27} erg sec), and f is the frequency. As shown in (1-1) above, $f = c/\lambda$, hence, $E = (hc)/\lambda$. With appropriate units, (hc) becomes 12.4.

This dual character of electromagnetic waves is foreign to our ordinary experience and was accepted generally only after the evidence was overwhelming. (This is but one of the many apparent contradictions of modern physics and perhaps awaits new concepts to provide a unifying theory.) Photon properties are not at all evident in radio waves but become apparent in light and dominant in x-rays. Let us illustrate photon effects with three examples:

1. *The photoelectric effect.* Both light and x-rays can directly ionize atoms by this mechanism. In all cases a certain minimum photon energy is required, characteristic of the atom involved. This photon energy corresponds exactly to the energy required to ionize that atom, as found by other methods.

To illustrate, consider a photoelectric cell operated by light, used to open the elevator door in an office building. Ordinarily, white light is used. If a *deep red* light is used, the photocell does not respond and the door closes, no matter how long or brightly the light shines. The substitution of a *blue* light (shorter wavelength and hence greater photon energy) causes it to operate the door almost instantly, even with a relatively *dim* light! (White light works because it contains all colors, including blue.) Were the photocell operated by wave action, *any* color should work so long as the wave amplitude (brightness) were great enough. But evidently these photons act *individually*, and several cannot cooperate to drive out a particular electron from an atom. Each photon must have enough energy (small enough wavelength) to do it alone, or it fails.

2. *Compton scatter*. All electromagnetic waves are scattered in all directions when they hit objects, in accordance with principles developed by Maxwell and Thomson during the late nineteenth century. This "classical" scattering is characterized by one common fact: the scattered waves are unchanged in the process. Thus, a monochromatic blue light has exactly the same color or wavelength after scatter as before, and a scattered 22 megacycle short wave radio signal is similarly unmodified.

In a brilliant series of experiments and analyses, A. H. Compton showed that during scatter some x-rays acquire *longer* wavelengths. The classical theory is unable to explain this change in wavelength. However, the change is completely explained, even predicted as to extent and amount, using the photon theory.

3. *Photonuclear interactions*. In more recent years, devices have been developed which produce photons of many MeV energy (extremely short wavelengths). These can initiate remarkable effects. For example, they can produce *material objects* (electrons) out of nothing but a *photon*, and eject particles from a nucleus (photodisintegration)! These events are unexplainable by classical wave theory but are clearly predicted by the photon (or quantum) theory.

Recapitulation

The photon is a real conceptual challenge, so perhaps it is useful to summarize what photons are and are not.

1. Photons are bundles of energy. Their properties in atomic interactions are determined uniquely by the amount of their

energy, which also characterizes their wavelength as observed in diffraction measurements. Just as atoms with the same atomic number are similar to each other, so are photons of the same energy and wavelength.

2. *But here the resemblance to atoms ends. For example:*
 - a. Photons cannot form combinations like atoms.
 - b. *Photons have no mass.* Also, they can travel only with the speed of light, and no slower; correspondingly, objects with mass cannot *attain* the speed of light.
 - c. Only 103 or so types of atoms are known. Photon energies are *infinite* in variety.
 - d. Actually, photons are simply radiant energy released from disturbed matter.
 - e. A photon can be *lowered* in energy by giving some to an object in the way *but never raised in energy.* Example: A 3 eV violet light photon is absorbed by a black surface.

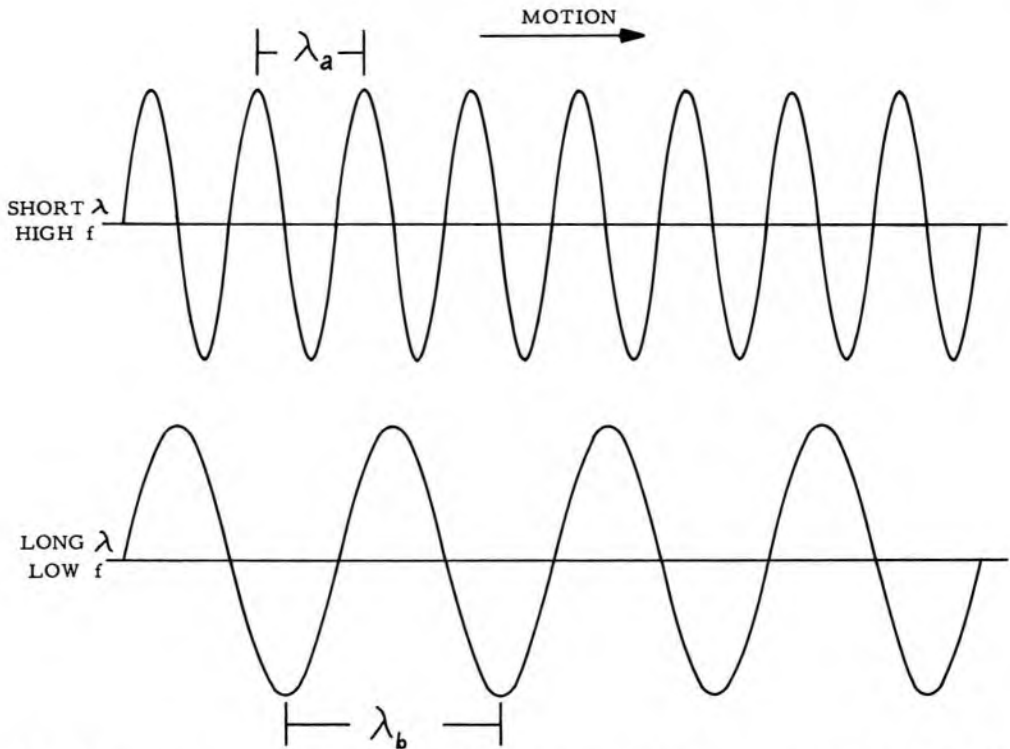


Fig. 11. Memory device for recalling some photon properties. A photon is pictured here as a train of waves whose peaks are separated by λ and whose amplitude and duration are the same for all kinds of photons. If these waves represent the electric fields, the upper wave will force an electron to oscillate (like the cork in Fig. 9) twice as many times as the lower wave. It therefore conveys twice as much energy to the electron. Note that its wavelength is *half* as great, consistent with $E = 12.4/\lambda$. (Only a few cycles are shown to facilitate drawing; actually, an enormous number is present even with light. Buildup and decay at the ends are also not shown.)

The surface acquired 3 eV of heat (atomic motion); the photon loses all its energy and ceases to exist.

In general, a photon has no way to go but *down* in energy—a depressing future.

Is there a way the dual character of photons can be simply considered? Photons are not basically simple, so the answer is probably not without gross distortion. Figure 11, a conceptual crutch often proved helpful, is reluctantly presented with the fervent admonition to lean on it lightly!

In this view, the photon is taken to be a train of waves whose peaks are separated by λ and whose amplitude and total duration are the same for all kinds of photons. Two photons are shown of long (B) and short (A) wavelengths.

Note λ is bigger, but f smaller in (B). Also, an electron is forced to vibrate fewer times by passage of this photon than by (A), so less energy is communicated to the electron. This all checks with the relationship $E = 12.4/\lambda$.

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2

Basic Electricity and X-ray Machines

Our review of atomic structure and electromagnetic waves has demonstrated their basically electrical nature. However, this review has relied to some extent upon the reader's general background and used without definition such terms as charge and electric field. A more formal presentation of these and other terms and concepts is required for comprehension of more advanced topics, such as measurement of radiation dose and radioactivity. Furthermore, a basic knowledge of electrical circuit principles is essential to understand the operation of x-ray machines and radioactivity instruments.

This chapter is divided into two parts: a review of some aspects of basic electricity and a discussion of the parts and basic circuits of x-ray machines. More advanced aspects of circuit components and design are beyond the scope of this book.

ELECTRICITY

This section considers electrical concepts, starting with terminology and then proceeding to current conduction and wave forms.

Basic Terms*

CHARGE. The ancient Greeks observed that when pieces of amber are rubbed with fur they acquire the ability to attract small pieces of light material like cork or pith. (The word electricity is derived from the Greek word for amber, *elektron*.) This effect differs from magnetism in that all materials can be electrified with proper technique.

Later experiments revealed many interesting facts about electrified objects, such as the following:

1. If one rubs hard rubber rods briskly with fur, they acquire the ability to repel each other strongly. Glass rods also repel each other similarly when rubbed with dry silk.

* See Table 1 (p. 36) for summary of useful electrical units.

2. Electrified glass and hard rubber rods always *attract* each other. Moreover, the electrified fur *attracts* rubber but repels glass rods; the electrified silk *attracts* glass but repels rubber rods.

It is evident that there must be two different ways of charging or electrifying objects. Franklin suggested that the rubber rod was charged with negative, and the glass rod, positive electricity. Experiments showed that combining positively and negatively charged objects tended to reduce or neutralize both charges.

We now believe that an electrified rubber rod has acquired extra electrons by direct transfer from fur, leaving the fur deficient in electrons. Similarly, glass can give up electrons to silk. More generally, gross objects are normally electrically neutral, with equal numbers of electrons and protons; they become charged negatively or positively, respectively, upon gaining electrons from, or losing them to, other objects. (Since electrons have mass, charges are always associated with matter. Note that electric charges differ basically in this regard from photons, which have *neither* charge nor mass.)

The electron's charge is relatively great for its tiny mass. For this reason, electrons are quite mobile and reactive. They are readily lost and acquired by atoms (and groups of atoms) to form ions and other charged objects of, respectively, positive and negative charge.

As indicated above, like-charged objects repel, and unlike attract each other. One might logically expect that the force between two charges increases proportionally as either charge is increased, and this is indeed true. For example, doubling one charge *doubles* the force with which it acts on the other. Doubling both charges *quadruples* the force. One might expect such forces to decrease when charges are separated. This decrease is also found but perhaps more rapidly than expected. The force is inversely proportional to the *square* of the distance, just as with gravitational and magnetic forces. Thus, tripling the separation of two tiny charged particles reduces their force on each other not three but *nine* times.

The electron charge is the natural unit in considering interaction of radiation with atoms and atomic interactions in general. However, a larger unit is needed for most measurements, such as defining the unit of x-ray exposure: the roentgen (R). For such purposes, the *electrostatic unit* (esu) is more convenient. This is the amount of charge, on a tiny object, which repels an identically charged tiny object 1 cm away in vacuum with a force of 1 dyne, about the weight of a milligram at sea level. More simply, it is equal to the charge of 2.08 billion electrons. For electrical equipment and machinery, an even larger unit, the *coulomb* (C), is used. This is 3 *billion* (3×10^9) esu's or 6.24 *billion-billion* (6.24×10^{18}) electron charges (Table 1).

TABLE 1. USEFUL ELECTRICAL UNITS

A. Electric Charge		
electronic charge	(e)	charge on an electron
* electrostatic unit	(esu)	2.08 billion electron charges
coulomb unit	(C)	3 billion esu, or 6.24×10^{18} (6.24 billion-billion) electron charges
B. Electrical Current		
ampere	(A)	1 coulomb/sec, or 6.24×10^{18} electrons/sec.
milliampere	(mA)	1/1,000 ampere (10^3 A)
microampere	(μ A)	1/1,000,000 ampere (10^{-6} A)
nanoampere	(nA)	1/1,000,000,000 ampere (10^{-9} A)
micro-microampere or picoampere	(pA)	1/1,000,000,000,000 ampere (10^{-12} A)
C. Electrical Voltage		
volt	(V)	The potential difference between two points when one joule of work is done against electrical forces in moving one coulomb of charge between these points.
kilovolt	(kV)	1,000 volts (10^3 V)
megavolt	(MV)	1,000,000 volts (10^6 V)
† beavolt	(BV)	1,000,000,000 volts (10^9 V)
D. Electrical Resistance of Impedance		
ohm	(Ω)	resistance of an international standard
kilohm	(k Ω)	1,000 Ω ($10^3 \Omega$)
megohm	(M Ω)	1,000,000 Ω ($10^6 \Omega$)

* Used in defining the roentgen.

† The preferred unit is "gigavolt," or GV, but BV continues to be widely used.

CURRENT. Electric current is the flow of electric charges, just as water current is the flow of water. The analogy is limited because water flows bodily but charges in liquids and solids travel in rather complex and peculiar ways. Pipes or similar devices confine water flow, with valves for controls. Electrical current in machines flows mainly through sheathed metal wires, usually copper, with switches for control.

We think of the current magnitude as the rate at which charges pass by. In principle, one should be able to count charges going by like anything else, such as automobiles on a highway, or people enter-

ing a department store. Several units of current are useful for various situations. The obvious one is electron per second, but the larger more practical unit is coulomb per second. This has been given its own name of *ampere* (A). For instrument and x-ray tube currents, smaller units 1/1,000th and 1/1,000,000th this size are more useful: these are called the *milliampere* (mA) and *microampere* (μA), respectively (Table 1).

CIRCUITS. Electrical currents can be free or confined, just as water flow can be free, as in rain or a flood, or confined to a river bed or pipes of a building. In electrical equipment and instruments, currents are almost always not only confined but also made to circulate in closed paths called "circuits." Graphical representations of circuits are called "circuit diagrams." In addition to the pathways (wires), these diagrams also symbolically show the electrical objects through which the current flows. There are three basic electrical components, called resistors, inductors (coils), and capacitors (condensers). In addition, circuit diagrams show various control devices (switches, vacuum tubes, and transistors) that turn currents on and off or adjust their magnitude, analogously to valves and pumps in plumbing systems, but with vastly greater speed and versatility.

Circuits all have one characteristic in common: at any instant, the current arriving at a point is the same as that leaving the point; charges cannot accumulate at a given spot. Figure 1 shows four basic circuit arrangements. In the simple circuit (1), voltage* V causes electrons to flow in the direction indicated, with current I. A characteristic of the circuit is that I has this same value at *all* points, no matter where one may test—including the source of V, or the object A.

In the simple series circuit (2), we have introduced other circuit

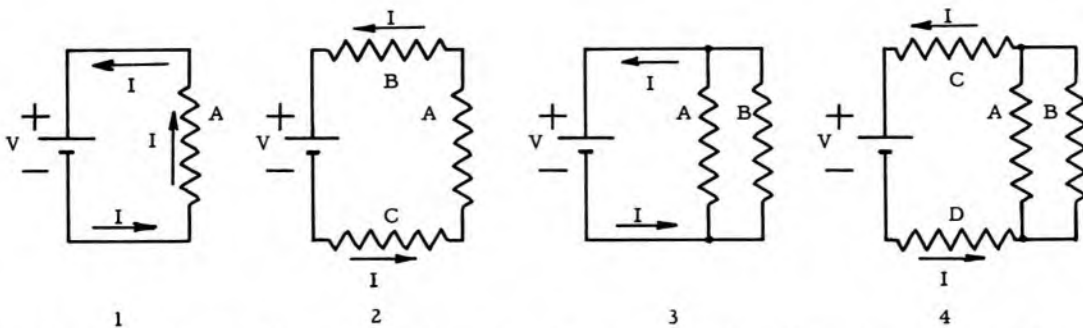


Fig. 1. Four basic electrical circuits. 1. Simple circuit—A only. 2. Simple series circuit—A, B and C in series. 3. Simple parallel circuit—A and B in parallel. 4. Series-parallel circuit—A and B in parallel, and their combination in series with C and D. (Two other circuits are not shown, wye and delta; they are of considerable engineering importance but beyond our interest in this book.)

* Voltage is defined on page 38.

elements in a "series" array, i.e., the same current goes through all these elements. Of course, even a simple circuit must have wires leading to the object (such as a line cord to a television set). The wires are circuit elements also, so we are always dealing with series circuits. However, the combinations can be made to approximate simple circuits by design, by using sufficiently heavy wires.

In the simple parallel circuit (3) two objects share a common voltage instead of current, and their graphical representation shows them parallel to each other. Note that the main current (supplied by the voltage source V) divides upon entering A and B and then is reconstituted on emerging from them. However, as before, at any instant the total currents entering and leaving the parallel pair are the same, since this is a circuit.

In practice, there is always some resistance in the circuit parts leading from V to the parallel elements. Figure 1 (4) shows this indicated as C and D , and the array is called a series-parallel circuit, having attributes of both circuit types.

While some circuit arrangements may be very complex, they can all be represented by combinations of series, parallel, and one other array (wye or delta) which is beyond the scope of this discussion.

VOLTAGE. Charges move from place to place, in general because they are caused to do so by forces from other charges. Since like charges repel, electrons tend to move from minus and toward plus charges, and protons or plus ions from plus toward minus charges. We speak of this motion as due to an electromotive force since it refers to a force tending to move electrons. In all paths some such motivating agency is needed to produce and maintain currents against the opposition of objects carrying them; this is referred to as voltage, whose unit is the volt.

Strictly speaking, electrical potential or voltage is not a force but an indication of how much energy is available to do work on each charge. It is conceptually useful to introduce a gravitation analogy. Consider a grand piano hanging from a cable outside a fifth-floor window (Figure 2). The piano at 45 feet above ground level has gravitational energy of position. This energy also depends on the weight involved. Obviously, the piano has more energy than one of its wheels, as snapping of the cable would demonstrate to a pedestrian below. We say both the wheel and the piano have the same gravitational potential (or height), but the actual energy also depends on the object's weight. The potential is really energy per unit weight involved.

Analogously, electrical potential corresponds to the potential energy per charge involved. Here, too, the energy arises from the fact

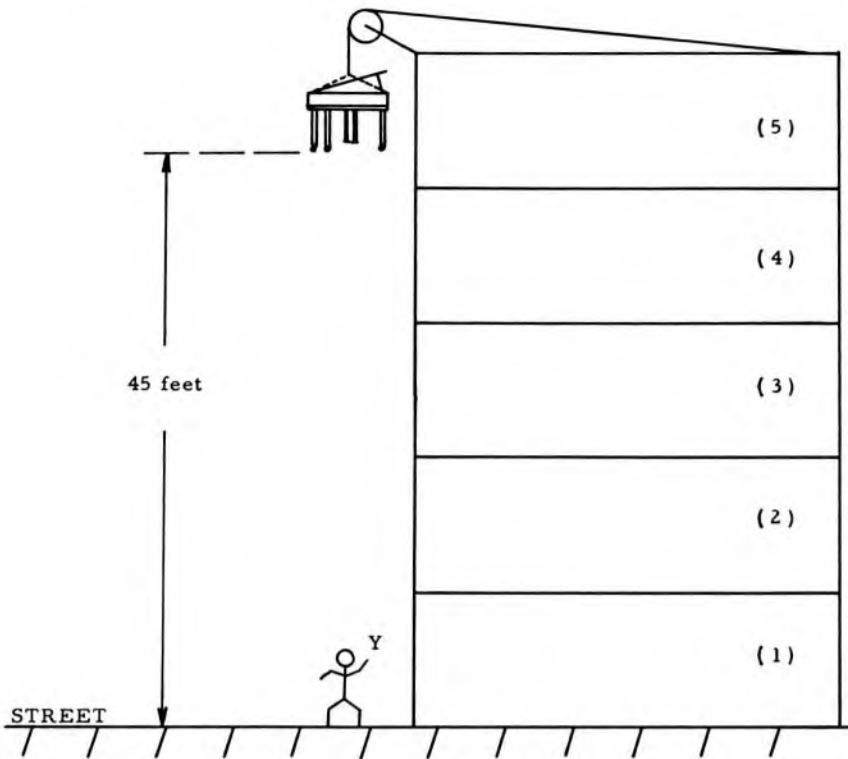


Fig. 2. Potential energy and potential. The piano has energy of position, as the rash pedestrian Y might discover too late if the rope breaks; so has the wheel on a piano leg. Of course, the wheel's energy is much less than that of the entire piano because it weighs less. Both are 45 feet above the ground and have the same potential, but the heavier object has more *potential energy*, corresponding to the greater work done in overcoming gravity when it was raised. Similar reasoning applies to electrical as to gravitational forces. In electricity, the potential energy is in electron volts (eV) and the potential in volts (V). The gravitationally relevant quantity is the object weight; in electricity, it is its electric charge.

that two locations have different *electrical potential*, but the more charges moved between these locations the more work is done and the more *energy* involved.

The piano's potential energy arises from the pull of one object (earth) on another (the piano) elevated above the ground. Electrical potential arises from one charge pulling or pushing any other brought near it. If we only see the piano fall (because of the earth's pull), or an electron move (because of another charge's action) we attribute this to unknown cause. One often uses the term "field of force" or "field" to describe such phenomena. For example, we can say that the piano fell because it was in the *gravitational field* of the earth, and the electron moved because it was in a corresponding *electric field*.

The unit of voltage is the volt, the potential difference which produces an ampere current in a standard one ohm resistance (Table

6).^{*} Less formally, it is roughly 1/12 the voltage developed by an automobile battery containing six lead cells. As a result, a coulomb of electrons leaving the minus terminal can do 12 joules' (12 volts \times 1 coulomb) worth of work in turning the engine starter motor.

In radiology, voltages can be rather large, and larger units are often useful (Table 1).

RESISTANCE AND IMPEDANCE. When currents flow through matter, the moving charges encounter charges of atoms and interact with them. This causes the atoms to move about more rapidly, producing heat. In this case, electrical energy has been dissipated *thermally*. It is analogous to the production of heat due to mechanical friction when one object slides over another. Sometimes this conversion to heat is desirable, as in an electrical heating appliance. In other cases, however, it is undesirable and one resorts to design to minimize it.

In a simple dc circuit (such as one involving a battery), resistance is the only factor limiting current flow. However, currents in some situations may be fluctuating or periodic (as the waves discussed above). (These "alternating currents," or ac, are discussed below.) The value of current flowing is then generally limited to lower values than the simple resistance alone would indicate. There are two reasons for this. Alternating current flow causes additional types of electrical energy conversion into heat, so that ac resistance is normally *greater* than that for dc; and, coils and capacitors oppose *changes* in current. This effect is called *reactance*, of which there are two kinds, due to inductance (as in coils) and capacitance (as in capacitors). The total opposition to flow of current results from the aggregate effect of all these factors (resistance, inductance, and capacitance) and is referred to as *impedance*. The electrical unit for both resistance and reactance, as well as impedance in general, is the ohm. An ohm is a resistance or impedance which limits the current from one volt to an ampere. A dramatic illustration of the difference between impedance and resistance is the consequence of applying dc voltage across a transformer. In general, the transformer may overheat and be damaged by a dc voltage of magnitude perfectly safe with ac.

Current Conduction

Let us now restrict ourselves to resistance since further consideration of reactance is beyond our concern in this discussion.

^{*} More correctly, the coulomb is defined as 2.9979×10^9 statcoulombs or esu's, which in turn is defined in terms of mutual repulsion between two point charges (p. 35). Then, one volt is the potential difference between two points when one joule of work is done against electrical forces when a coulomb of charge is moved between these points.

There is a wide range of ability of different materials to transport current. Conductors carry currents readily, insulators minimally, and semiconductors in between. The orders of magnitude of currents usually carried by them are, respectively: amperes (A), picoamperes (pA), and milliamperes (mA).

Conductors are almost invariably metals, in the ordinary range of temperatures. Semiconductors include both water solutions and solid non-metals and some of their compounds. The electrical properties of such solids are critically dependent on the nature and type of any trace impurities present and on the perfection of their crystalline structure. Finally, insulators include various natural and synthetic organic and silicon compounds. Examples of various materials employed in all three categories are given below.

CONDUCTORS. Perhaps the most important conductors are cylindrical wires, almost always made of electrolytic copper. They are generally sheathed in appropriate insulators to prevent undesired connections to adjacent conductors. Examples of other conductors in x-ray machines are:

1. *Switches*—Movable conductors of copper alloys; also electronic switches.
2. *Fuses*—Bismuth and lead alloys which melt when carrying excessive currents, interrupting the current and thereby protecting associated circuit components.
3. *Solder*—Tin and lead alloy, usually with its own rosin flux, provided to reduce oxides produced during soldering. Solder is used to render connections between conductors permanent.
4. *Vacuum tube electrode materials*—Tungsten, molybdenum, kovar, silver, gold, etc.

INSULATORS. Insulators are used to prevent undesired connections between different electrical circuits. Examples of their uses are:

1. *As insulating sheaths for wires.* Because it deteriorates with time, ordinary rubber has largely been replaced by various thermoplastics of the vinyl, polyethylene and teflon families. Some thermosetting plastics also give a tough varnish coat to wires used in winding coils.
2. *For circuit panels and switches, to support terminals and components.* Bakelite and similar thermosetting plastics are widely used for this. Special switches may require other materials, such as ceramics.
3. *High voltage insulation.* Ordinary voltages are only a few hundred volts different from that of objects and people in an x-ray room and are readily insulated against. However, x-ray

generators regularly operate between 20,000 and 150,000 and more volts from ground and tend to produce undesired currents to patients and operators as well as to metal supports and containers. These can result in injuries or machine damage if care is not taken in design and operation. Even air insulation can break down at a few thousand volts if conductor terminations are sharp and too near each other. Most modern x-ray equipment therefore employs high grade insulating oil to surround both the x-ray tube in its metal container, and the high voltage generating unit in its metal box. X-ray and valve tubes are made of Pyrex glass, with a very good vacuum inside. Failure of such tubes can occur if excessive amounts of gas accumulate in the tube because gases can become conductive at high voltage. The high voltage cables are insulated with good quality flexible plastic; terminals are of high quality Bakelite material.

4. *Instrument type insulators.* For ionization chamber instruments such as the familiar Victoreen unit, polystyrene and amber are used because they are very resistant to surface and volume accumulation of water. Silk and most other natural fibers are poor insulators unless very dry. Fused quartz and similar glasses are excellent insulators but difficult to fabricate.

SEMICONDUCTORS. Semiconductors have many radiologic uses. Graphite is used to conduct collected charges in ionization chambers. Cadmium sulphide crystals act as radiation detectors since their electrical resistance decreases markedly under irradiation. Tiny gallium arsenide and silicon crystals, appropriately processed, have been used as "p-n" junctions for detection of corpuscular radiation. Silicon rectifiers will probably ultimately replace valve tubes in x-ray machines, and single semiconductor crystals of various kinds are used in laser beam units.

OHM'S LAW. The current through an object normally increases with the applied voltage. In practice, it is found that over a fairly wide range there is a simple, proportional relationship in many materials, that is,

$$I = V/R \qquad (2-1)$$

The quantity R is called the resistance, when the current is simple dc (see below). (For ac, the *impedance* must be used, of conventional symbol Z.) When V is in volts and I in amperes, R must be in ohms.

Wave Form

In most circuits the current magnitude varies with time. This variation is described by the wave form, which is a graph of magnitude vs. time for current or voltage. Figure 3 shows some wave forms useful in x-ray work.

dc is short for direct current, ac for alternating current. dc is current which may vary in strength but does not reverse in direction. Constant current is a special kind of dc. Figure 3 (1) shows the wave form of a constant voltage. As provided by a battery or a special x-ray

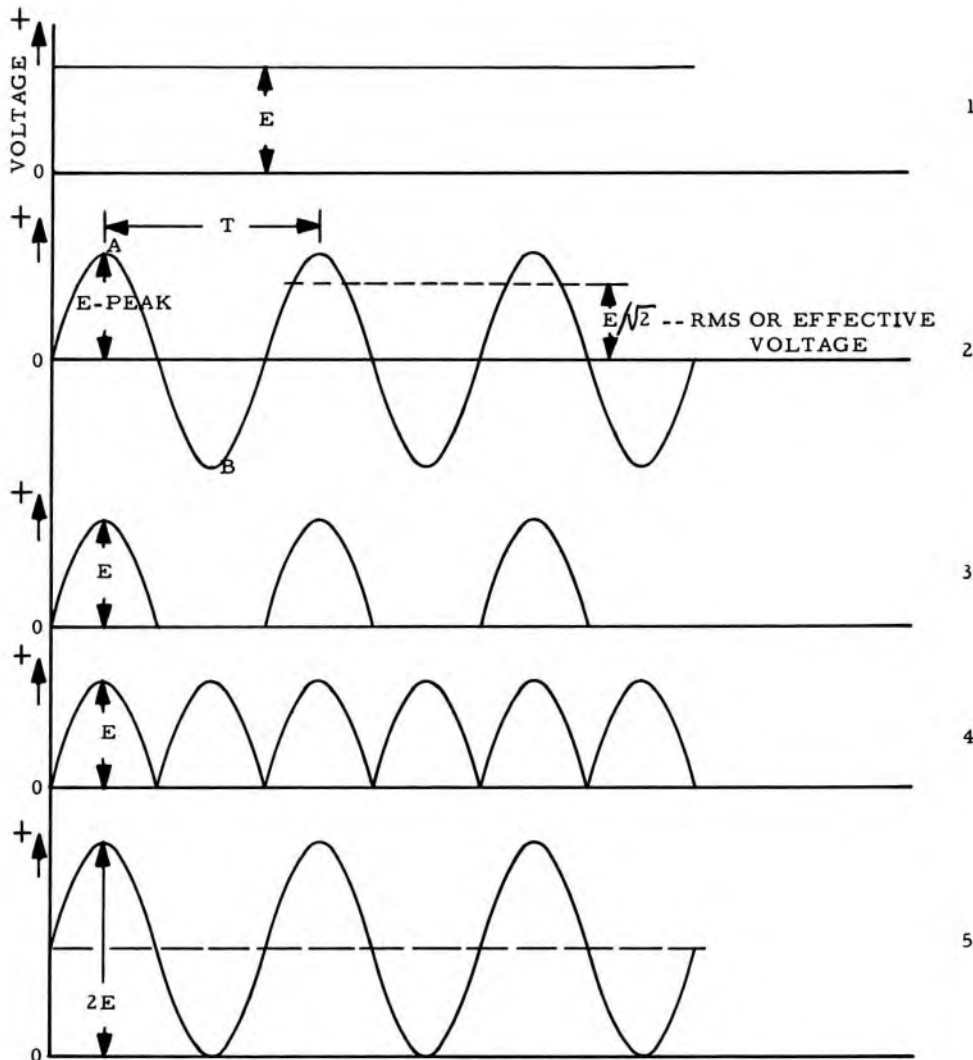


Fig. 3. Useful wave forms in radiology. 1. Simple dc voltage—constant potential. 2. Simple ac voltage. The terms: T —period, the time between successive maximum values of the same polarity (time between crests); E —peak voltage; $E/\sqrt{2}$ —effective value of voltage; also called RMS. 3. Pulsating dc voltage—half wave. 4. Pulsating dc voltage—full wave. 5. Pulsating dc voltage—Villard. The above wave forms are idealized and may be greatly distorted in practice.

machine circuit it remains not only unidirectional but substantially constant in value. More frequently, x-ray tube currents are *rectified* currents. While these are dc, they pulsate between approximately zero and a maximum value [Figs. 3 (3), (4), and (5)].

Current supplied by the electric company is almost universally ac because this is easier than dc to produce and transfer from place to place. It is a vibrating current. Vibration is a normal state of matter. Mechanical vibration is found wherever power machinery operates, as in the automobile, train, elevator, or washing machine. Smooth repetitive motion is described as "sine wave" or sinusoidal in character, which is closely related to circular motion. Alternating current is also basically sinusoidal in character [Fig. 3 (2)]. An inherent advantage of ac over dc power is that the magnitude of ac voltages can be readily increased or decreased by large factors, by employing a simple very convenient device called a transformer, described at some length on page 59.

We shall refer below specifically to the wave forms of Figure 3 in discussing rectification and x-ray machine operation.

SOME TERMS DEFINED. Certain terms are commonly used in discussing ac. Refer to Figure 3 (2).

1. *Period.* This is the time interval between successive crests of the ac wave (T).
2. *Frequency.* This refers to the number of times per second the voltage reaches a maximum of a certain polarity (the number of crests each second). For example, ordinary power company currents are 60 cycles per second in most parts of the United States. The corresponding period is 1/60 of a second.
3. *Maximum value of the voltage.* As shown in Figure 3(2), the voltage rises from zero to a maximum value at A, decreases to zero, rises to a maximum value of opposite polarity at B, then repeats the process. The magnitude of the highest value of voltage is the *maximum* or *peak value* (E). In x-ray work, this is one of the most significant quantities in determining the x-ray output from an x-ray tube.
4. *Effective or RMS value of the voltage.* The maximum is attained only momentarily; the rest of the time the applied voltage is lower and is actually zero twice each cycle. Hence, the energy delivered to a customer by the electric company cannot be honestly calculated from the peak value. For such purposes, the effective or root mean square (RMS) value is more useful. This is defined as the value of a constant voltage or current of the same appliance which would yield the same heating effect as the ac in question in the same heating appliance. It is actually $1/\sqrt{2}$ times the peak value (0.707).

THE X-RAY MACHINE

X-rays are produced when high-speed electrons collide abruptly with an object. In x-ray tubes, the electrons are supplied by a coiled wire called a "filament" (Fig. 4), which is heated to incandescence in a well-evacuated glass container by passing an electric current through it (similar to the filament in an incandescent electric light bulb). Just as heating water boils off water vapor, heating a metal boils off some of its electrons by giving them enough energy to escape the surface. The hotter the filament, the faster the kettle boils; for essentially the same reason, the hotter the metal surface the more copious the supply of electrons. We use the adjustable separate current supply to heat the filament, in order to control this electron flow, and thereby the rate of production of x-rays.

To make x-rays, the electrons must first acquire high speeds. This is accomplished by impressing a high voltage between the filament and the anode. This voltage numerically equals the maximum acquired electron energy in electron volts, by definition of the electron volt. For example, if 80 kV is applied, each electron falls through a poten-

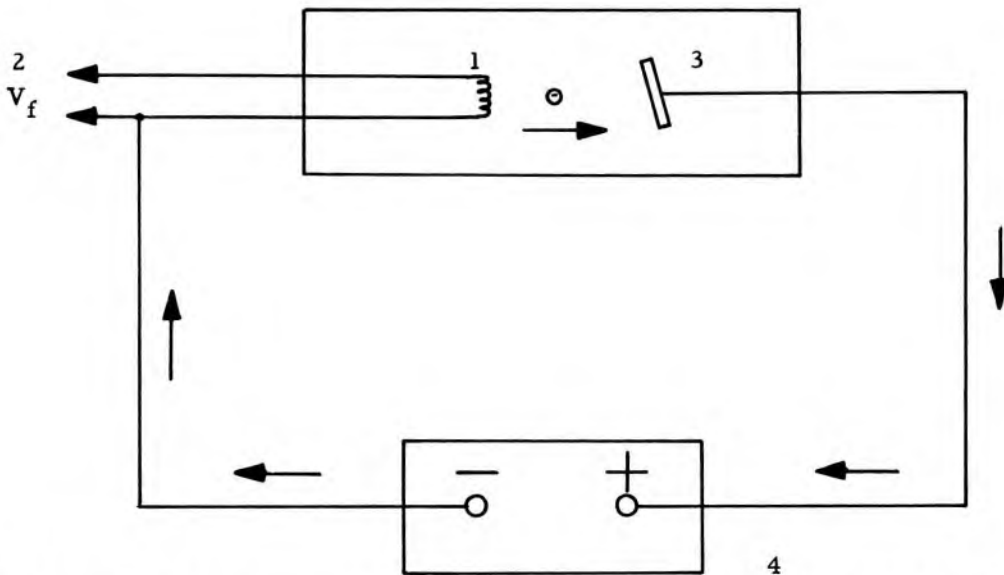


Fig. 4. Setup for producing x-rays. The x-ray tube includes the following: 1. A small coiled wire (filament) which is heated to incandescence and supplies the electrons. 2. The filament is heated by a current produced using an adjustable ac voltage. As V_f increases, so does the electron supply. 3. The electrons are accelerated to the anode (3) by an impressed high voltage supply (4) of 5,000 to millions of volts. The speed at impact depends on the voltage. Thus, 80 kV yields 80 keV electrons. A vacuum is needed in the glass tube, or the electrons lose their energy uselessly and possibly destructively in ionizing gas molecules.

tial drop of 80 kV, and by definition acquires 80 keV of energy.* In some cases, pulsating dc is employed, and one then usually refers to the maximum or peak kilovoltage (pkV or kVp) employed (i.e., 80 pkV yields 80 keV peak electron energy).

The user of the x-rays normally desires them for a specific purpose, such as obtaining a radiograph, or irradiating a patient or preparation. He is very interested in the kind and amount of x-rays produced, usually respectively referred to as the quality and quantity delivered.

The *quality* of x-rays refers to their penetrating ability. Generally speaking, x-rays generated at higher pkV values penetrate objects more readily than those produced at lower pkV.†

The *quantity* of x-rays refers to the total x-ray energy produced each second by the tube. This also increases when the pkV is raised. However, if one wishes to increase the quantity without substantially affecting the quality, the tube current (mA) or the exposure time or both may be increased.

In summary, the x-ray machine must provide the following items to operate the x-ray tube in a desired manner:

1. A conveniently and accurately adjustable separate current to controllably heat the x-ray tube filament which supplies electrons to produce x-rays.
2. An accurately adjustable high voltage to accelerate these electrons sufficiently to produce x-rays of the type desired.
3. Some sort of timing system so x-rays are produced for the desired length of time only.

The following discussion is intended to give a general idea of the essential parts of an x-ray machine and their functions.

The Main Units of an X-ray Machine

Figure 5 shows a block diagram of the four basic units.

The control panel enables the operator to select the pkV, tube current, and duration of the x-ray exposure. In addition, safety controls and meters are included. The modern panel is supplied with many control devices which perform various required adjustments automatically.

The transformer unit contains transformers which produce high voltage and directly energize the x-ray filament. These are called the high tension and x-ray filament transformers, respectively. In addition, high voltage rectifiers are also included in most machines, to

* This assumes the electron does not lose energy on the way to gas molecules as in a gassy tube.

† Peak kilovoltage is often alternatively referred to as kVp, rather than pkV, the convention followed in this book.

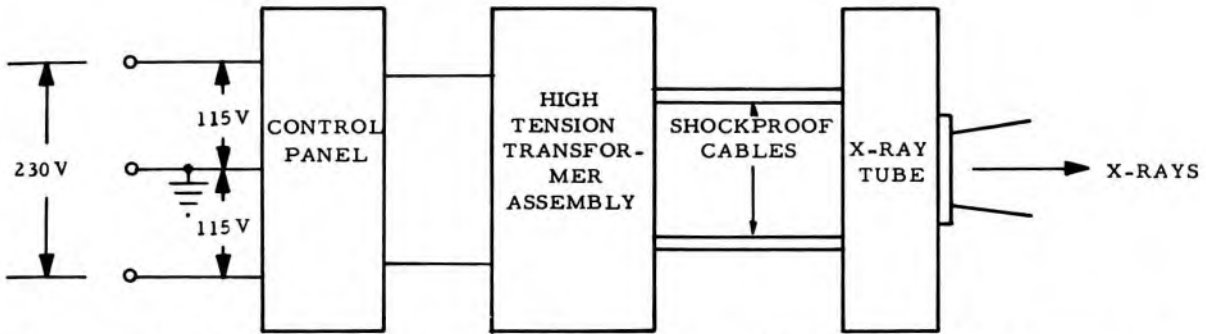


Fig. 5. The four basic parts of the x-ray machine. The control panel is used to select kV and mA; the high tension transfer unit produces high voltage and generally rectifies it; shockproof cables safely communicate this voltage to the tube; and the tube makes x-rays. A 230 V supply is normally employed in larger machines and is so indicated in the drawing.

produce dc high voltage for the x-ray tube from the transformer ac high voltage. A major benefit from use of rectifier units is that they enable x-ray tubes to operate safely at much greater mA values than would otherwise be possible.

Since x-ray tubes operate at potentially lethal high voltages, special "shockproof" cables have been developed to connect high voltage to the tube assembly. Incidentally, both the tube assembly and the transformer unit are also of "shockproof" design. This makes it permissible for a patient to touch the cable, transformer box or x-ray tube housing during an exposure without significant shock hazard. (It is generally bad practice for the operator to be in a room with x-rays turned on, but primarily for radiation, rather than electrical hazard, with modern machines.)

The x-ray tube assembly contains the x-ray tube, which is immersed in oil for electrical insulation. Within the tube an electron current beam of adjustable magnitude is directed at a metal target and strikes it with energies determined by the pkV employed.

Specific Parts of the X-ray Machine

As with automobiles and other expensive machinery, x-ray machines may range from simple to very elaborate. However, in all units the control panel must provide means to select exposure factors, protect the machine from damage to expensive parts, and operate with safety to patients and personnel.

CONTROL PANEL SELECTOR SWITCHES. Switches are employed to permit convenient choice of pkV. Usually there are two controls: a "major" dial for larger steps (usually 10 pkV) and a "minor" dial with steps of 1 or 2 pkV, but many machines use a more complex single dial. However, the selected value will be achieved only if the appro-

ropriate voltage has been supplied to the high tension transformer by the selector switch. Unfortunately, the incoming supply voltage to the machine may vary over a wide range from causes originating both inside and outside the building. This produces corresponding pkV variations. For example, a 230 V supply line may drop to 207 V in a building when heavy power machinery starts. The x-ray tube voltage is then correspondingly lowered, perhaps to 81 pkV instead of the 90 pkV for which the dial was set, and the resulting roentgenogram is very light.

To avoid this difficulty, an additional separate switching system is provided to permit compensation by the operator for line fluctuations. Before making the exposure the operator first adjusts a switch until the proper voltage is supplied to the pkV selector circuit. The switch does this by varying the position on the autotransformer (see p. 64) to which the supply voltage is connected.*

In diagnostic x-ray machines an mA switch is usually provided for selecting certain cardinal values, such as 5, 25, 50, 200, 300, and 500 mA. In other applications an exact control of mA values is required and continuous adjustment and good mA meters are provided (in some industrial and in all therapy applications). In all cases, what is varied directly is a current controlling device such as a variable resistance or a solenoid reactor. (The former can be either a switch arrangement or a continuously adjustable resistor called a "rheostat." The solenoid reactor is a variable inductance unit which increases in *impedance* as iron is inserted into a coil, thereby reducing current in much the same way as an increase in resistance would. Such reactors are also used to dim lights in a theater.)

CONTROL PANEL—TIMER. Timer units are usually set manually to the desired time and the exposure started. At the preset time, the exposure is terminated. For times in the range of hours, minutes, and seconds, synchronous clock units with special switch arrangements are adequate. However, for exposure times less than one second (to 1/120 second) electronic timers are usually employed. (Grid-controlled x-ray tubes are best for extremely short exposure times if close reproducibility is required.)

The timer itself is ordinarily a rather low-power device whose switch can safely carry only a relatively small current. On the other hand, currents supplied to the high voltage transformer may be of the order of 5 to 60 amperes (rather high values). For this reason, the actual connection is made by a heavy duty switch called a "contactor"

* The pkV compensation cannot correct for rapid line voltage fluctuations. It is usually necessary to provide separate electrical supply transformers for x-ray installations to minimize such fluctuations.

or by powerful electronic switch tubes. These special switches are activated by low current signals from the timer and exposure switch circuit. When the timer turns off this *small* current, the contactor is designed to interrupt very quickly the large current to the high voltage transformer.

CONTROL PANEL—PROTECTION. Fuses and special relays called “circuit breakers” turn off current when it exceeds permissible values. A fuse is a replaceable unit substituting for a small portion of a wire. It readily carries the full current until it overheats and melts or “fuses,” thereby interrupting the circuit and protecting the machine from overload. This interruption is arranged by the fuse manufacturer to take place only when a predetermined permissible current level is exceeded (say 15, 20, 30, or 60 amperes for high power work). A circuit breaker is a switch normally kept closed by a spring. This switch carries the main current. Also carrying this current is a coil which can operate an electromagnet (see p. 60). The current makes the electromagnet pull against the spring tension but normally not hard enough to overcome it. However, when the current exceeds a predetermined value, the electromagnet pull overcomes the spring tension and opens the switch. The unit uses a mechanical latch so that once opened it must be manually closed (“reset”).

In case an x-ray circuit is interrupted, never replace a fuse or reset a circuit breaker without first checking and correcting the cause of the overload. Otherwise, marginal operation of the machine may damage expensive high voltage components like the x-ray tube, shock-proof cables, or high tension transformer.

Interlock and other safety switches are connected in series with the timer to make sure various important preconditions have been met before the x-rays can be turned on. In diagnostic work, such switches assure that the x-ray tube rotor, moving grids, and body section devices are set. In radiation therapy and industrial work, interlocks assure that the tube shutter is closed to start, the filter present is the one desired, tube temperature is safe, room door is closed, starting voltage is not excessive, etc.

THE TRANSFORMER UNIT. As mentioned above, the transformer unit provides high voltage and filament voltages. The high voltage is used to speed up the electrons striking the x-ray tube target. The filament voltage produces current which heats the filament to incandescence to supply the electrons. The filament is operated at low voltage but rather high current (about 3 to 12 volts, 3.0 through 5.5 amperes, respectively).

The filament transformer performs two essential functions. First, it conveniently provides the required 3 to 12 volts when supplied with

about 20 to 80 volts from the control panel circuit, and second, it effectively isolates the x-ray tube filament from the control panel. This is essential since the filament as a whole is maintained at high voltage with respect to ground because of the way it is connected to the high voltage supply* (actually about half the operating pkV as shown below).

The high tension transformer produces high ac voltage from low. Voltage values obtained from the control panel (pkV selector switches) may range from 60 to 280 volts. They are used to derive voltages about 300 times greater for making x-rays. No simple alternative way has yet been devised to *increase* voltages.

RECTIFICATION—SELF-RECTIFIED CIRCUIT. Rectification is the process of obtaining direct current from an ac voltage. It will be recalled that most objects produce ac when ac voltage is impressed across them. This is true, for example, in a toaster, iron, or almost any other ordinary appliance. Most conductors do not rectify. (However, semiconductors under most conditions tend to produce some direct current when an ac voltage is impressed across them.) An x-ray tube itself will act as a rectifier under certain conditions. Figure 6 shows the circuit involved when an x-ray tube is connected directly across the high voltage winding of a high tension transformer. Under these circumstances, ac voltage (V) is impressed across the tube, but dc (I) flows through it (bottom). The filament can supply copious electrons which flow (arrows) when the anode is made positive with respect to the filament. However, the anode (under the conditions of operation favorable to this type of circuit) operates relatively cool. As a result, it cannot supply electrons to flow in the opposite direction when the voltage reverses in polarity—that is, when the filament is plus and the anode minus.

Unfortunately, in operation at great mA values a small part of the anode struck directly by the electrons may be raised to extremely high temperatures just short of the melting point of tungsten. Under such circumstances, the very hot tungsten of the target also serves as a source of electrons. During the time when the anode is negative and the filament positive, these electrons are driven to the filament and can cause it to melt, thereby destroying the tube.

Two basic considerations affect how hot the target gets. The first is the pkV and mA at which the tube is operated. If these are excessive, heat is poured into the target too fast, and its temperature rises too high. The second is the target area hit by the electrons. If this is too

* "Ground" is a term referring to the potential likely present in the x-ray room: of radiators, water lines, wet floors, windows, and other objects to which high voltage parts of the machine might spark over. For high voltage, shoes may provide little insulation, so people also are essentially at high voltage ground.

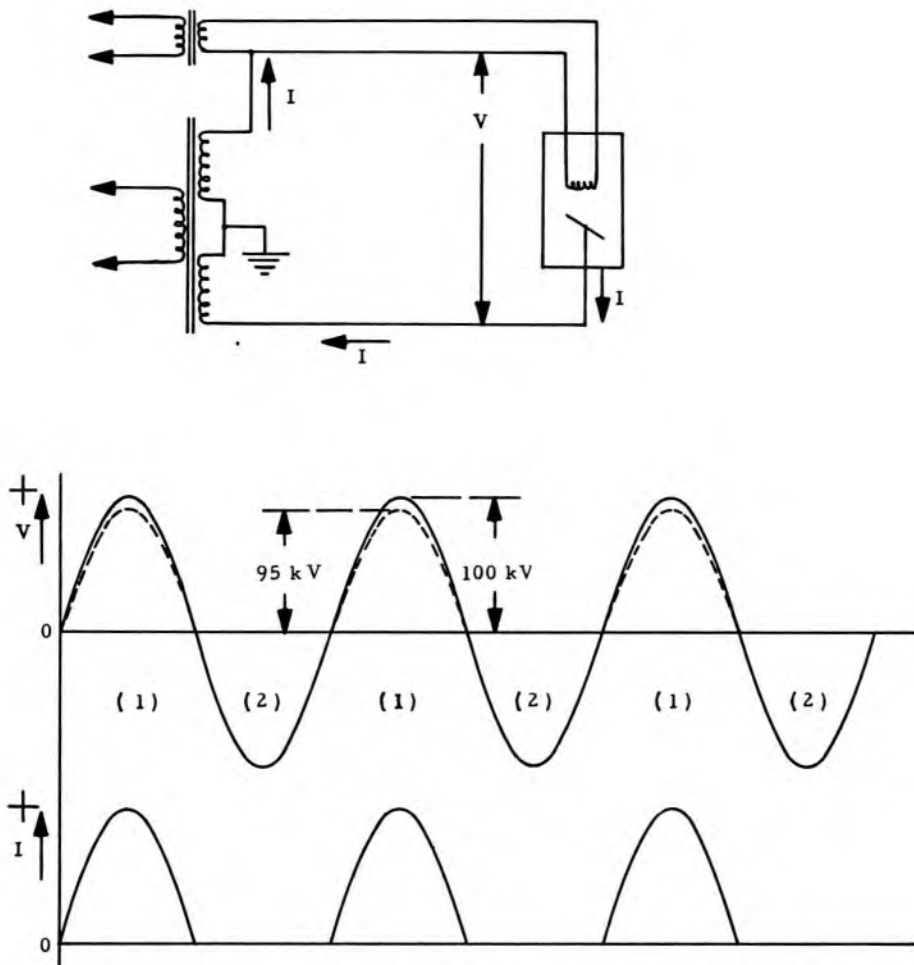


Fig. 6. The self-rectified x-ray circuit. Top. Note the high voltage ac is impressed directly across the tube. Bottom. Voltage and currents in the circuit. The lower curve shows the current flow. The solid upper curve shows the voltage produced by the transformer. This actually appears across the tube during the (2) half of the cycle, undiminished because the current is then zero. When current flows, however, part of the available kV is needed to overcome resistance of the transformer secondary winding. As a result, a lower pkV actually appears across the tube during (1), shown by the dotted line. For example, with a 50,000 ohm transformer resistance and 100 mA peak current, the peak tube kilovoltage is reduced 5 kV when tube current flows.

tiny, a blow-torch effect results in extremely high local target temperatures, even with only moderate pkV and mA values. These conditions severely limit the practical application of the "self-rectified circuit," as this arrangement is called.

In diagnosis, self-rectified circuits are therefore generally restricted to portable, fluoroscopic, and dental units. Self-rectified circuits are also employed in certain deep therapy machines, using special x-ray tubes capable of withstanding large reverse voltages. However, in almost all other applications a separate rectified circuit is employed

between the high tension transformer and the x-ray tube. In addition to preventing the destruction of the tube by inverse current as described above, rectifier circuits are also used to prevent damage to insulating structures such as the shockproof cables and the tube housing. They do so by preventing the impressing of the "inverse voltage" on these structures (Fig. 6). This inverse voltage may be of the order of 5 to 10 kilovolts greater than the voltage actually present while x-rays are produced and hence constitutes a more difficult insulation problem. The use of a separate rectifier confines this inverse voltage to the high tension transformer, or prevents it altogether.

RECTIFIERS. Two types of rectifying elements are employed in radiology. The more recently developed "solid state" type is generally made of silicon and vacuum tube rectifiers. Solid-state rectifiers have two inherent advantages: virtually indefinite life, and no need for filament voltage supplies to operate them. They are undoubtedly the x-ray machine rectifiers of the future. The more traditional vacuum tube rectifier units are called valve tubes, or diode rectifiers.

Valve tubes are similar to x-ray tubes in their basic components: filament and anode structures contained in an evacuated glass envelope. It is therefore of interest to consider why they should differ from x-ray tubes in their ability to withstand reverse voltages. The answer lies both in their construction and operation. First, valve tubes are designed so that electrons strike a much larger area of the anode and therefore produce less intense local heating. In addition, valve tube filaments are operated at considerably higher temperatures than x-ray tube filaments. This is accomplished by impressing much greater power* on them (of the order of twice as much as x-ray tube filaments). As a result, a valve tube during operation carries its maximum current with only about 250 to 1,000 volts required to move the electrons from filament to anode.

RECTIFIER CIRCUITS. Figure 7 shows a rectification arrangement employing two valve tubes in series with an x-ray tube. Consider the situation when the voltage is of such polarity that current flows through both the valve tubes and the x-ray tube (polarity "a"). The current is the same in all three tubes since this is a series circuit. However, the x-ray tube has rather high resistance; this is because it has a limited electron supply since its filament is operated at relatively low temperature. The valve tubes, on the other hand, have copious electron supplies and consequently offer low resistance. As a result, even with 100,000 volts across the x-ray tube, the valve tubes each have something like only 250 volts across them.

* Power refers to the joules per second (watts) dissipated in an object carrying current. It is calculated as volts \times amperes (effective values).

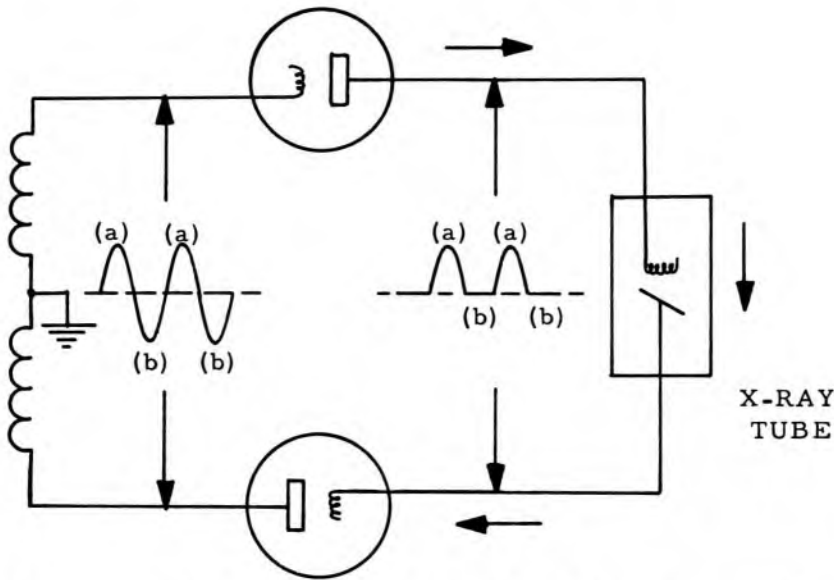


Fig. 7. Half-wave rectified circuit. Two valve tubes protect the x-ray tube and shockproof cable from the inverse voltage, permitting a higher operating mA. However, as shown, the current still flows only half the time, limiting the available x-ray output.

The net result is that the x-ray tube anode dissipates a power roughly 4,000 times greater than that of each valve tube's anode. For this reason, the valve tube anodes are much cooler than that of the x-ray tube. Now consider what happens when the polarity reverses (b).

Were the voltage now impressed across the x-ray tube directly, destructive reverse current might result. However, in the circuit shown each valve tube has a cool anode and hence cannot supply electrons to carry the inverse current. As a result, the valve tubes act like open switches and effectively disconnect the x-ray tube and shockproof cables during this half-cycle. When the polarity reverses and is appropriate for proper operation, the valve tubes again conduct.

The circuit referred to above with the two valve tubes produces a "half wave" wave form [Fig. 3(3)]. This is suitable where operating mA values are intermediate in magnitude, such as in some 140 pkV therapy machines and some light-duty diagnostic machines. However, in modern x-ray diagnostic work currents of the order of 200 to 500 and even 1,000 milliamperes are often required for good radiographic quality. For such purposes it is very important for the x-ray tube to conduct current during both parts of the cycle rather than only half the time. A more effective circuit is hence the 4-valve full-wave type, shown in Figure 8.

PEAK DETECTOR CIRCUITS. In 200 to 250 kV therapy, "Villard" and constant potential circuits are frequently used. These employ more complex rectifier circuits, with circuit elements called "capacitors."

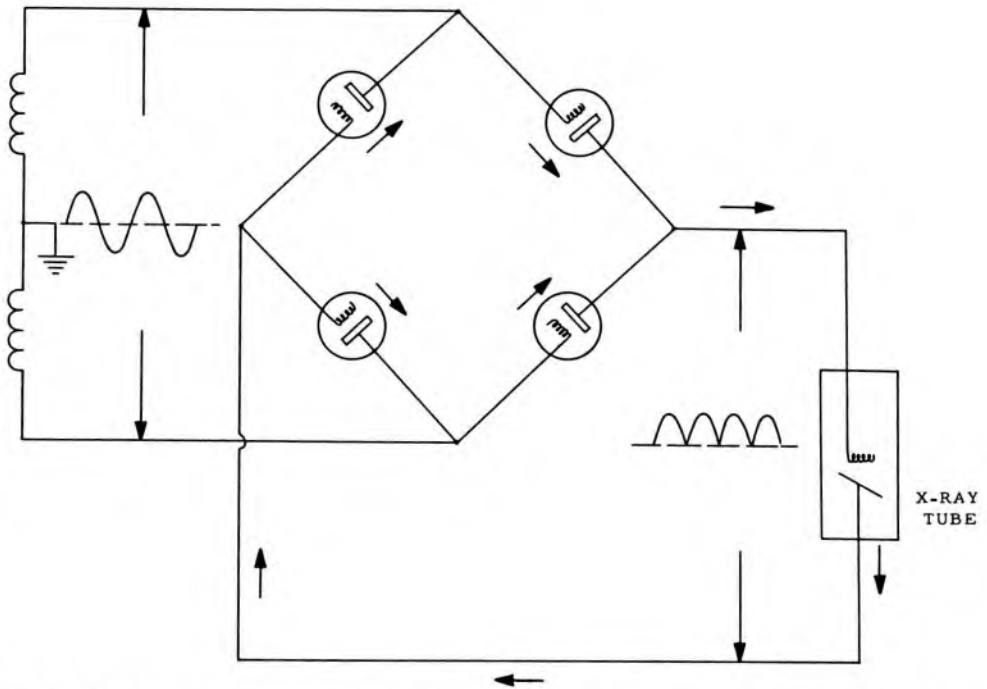


Fig. 8. Full-wave rectified circuit. Four valve tubes convert the ac voltage into a full-wave dc voltage. Hence, the tube can operate at higher average mA values, and the x-ray output is higher than with half-wave rectification. If one tube fails, the system becomes a half-wave circuit, but should be repaired at once to avoid accidentally exceeding system ratings.

Capacitors, often referred to as “condensers,” have the ability to store electric charges. This enables one to produce more constant dc voltage from pulsating voltage. Figure 9 shows the basic rectifier circuit employing a capacitor (the “peak detector” circuit), used in both of the therapy rectifier circuits. This consists of the high ac voltage supply, rectifier tube, and capacitor connected in series. The action of the rectifier is to permit current from the transformer to flow only during one-half the cycle. The capacitor accumulates charge until it reaches the maximum voltage of the transformer winding. Thus, a 100 pkV voltage developed by the transformer results in appearance 100 kV constant potential across the capacitor; when the capacitor voltage is less than this, some point is reached during the successive (a) cycles when the rectifier passes current to the capacitor from the transformer winding until the peak voltage is reached. The capacitor is prevented from discharging or losing its charge during the cycle (b) because the rectifier voltage is then of the wrong polarity for current to flow. Thus, the rectifier acts as an electrical valve both to raise the condenser voltage to equal the peak voltage and to prevent its discharge later in the cycle.

One of the basic principles of series circuits is that the voltage across any two components at any time is the sum of the separate

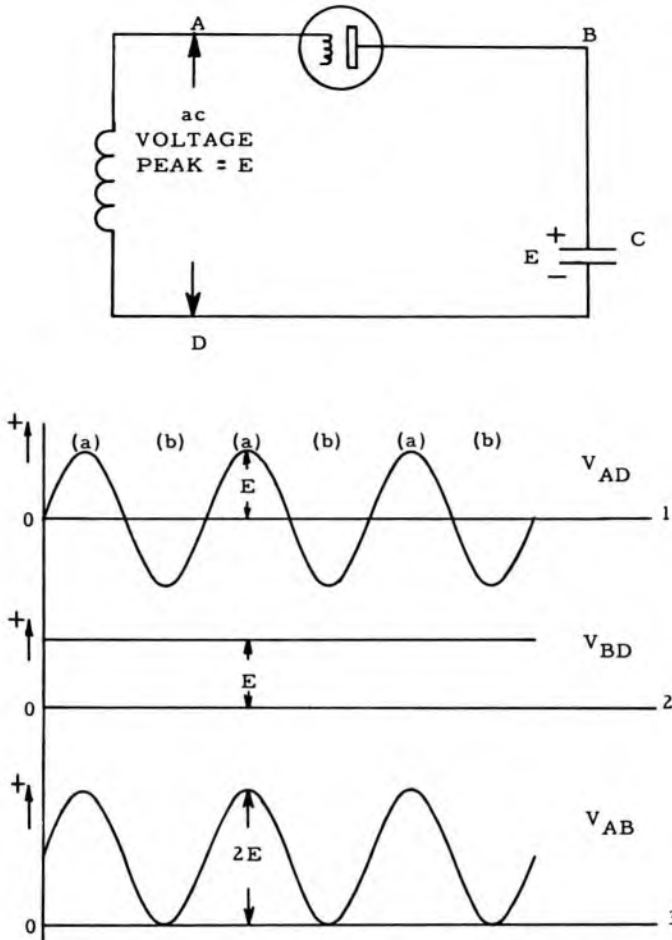


Fig. 9. Top. The basic peak-detector circuit. Bottom. Voltage wave forms: 1. ac voltage from transformer— V_{AD} of peak value E . 2. dc constant voltage across capacitor C — $V_{BD} = E$. 3. Pulsating dc voltage, of peak value double the magnitude of V_{AD} . The voltage appearing across valve tube is V_{AB} ; the peak value is $2E$.

voltages, just as currents add in parallel circuits. Advantage is taken of this fact in both constant potential and Villard circuits. The voltage between the points A and B in Figure 9 (Top) is the sum of the capacitor dc and the transformer ac voltages. During the appropriate part of the ac cycle, this voltage reaches a maximum *double* the value of the peak voltage itself! This is the basis for so-called “doubling” or more generally “multiplying” circuits. Alternatively, one can employ the constant dc capacitor voltage alone.

Figure 10 (Top) shows the use of two units of Figure 9 back-to-back to yield a constant potential. Note that the dc voltage achieved here is no greater than that provided by the transformer. The Villard circuit (Bottom) also consists of two basic peak detector circuits back-to-back, but arranged in such a manner that *four* voltages rather than *two* are added, yielding *twice* the transformer pkV. Figure 11 shows a more sophisticated constant potential circuit, consisting of a Villard

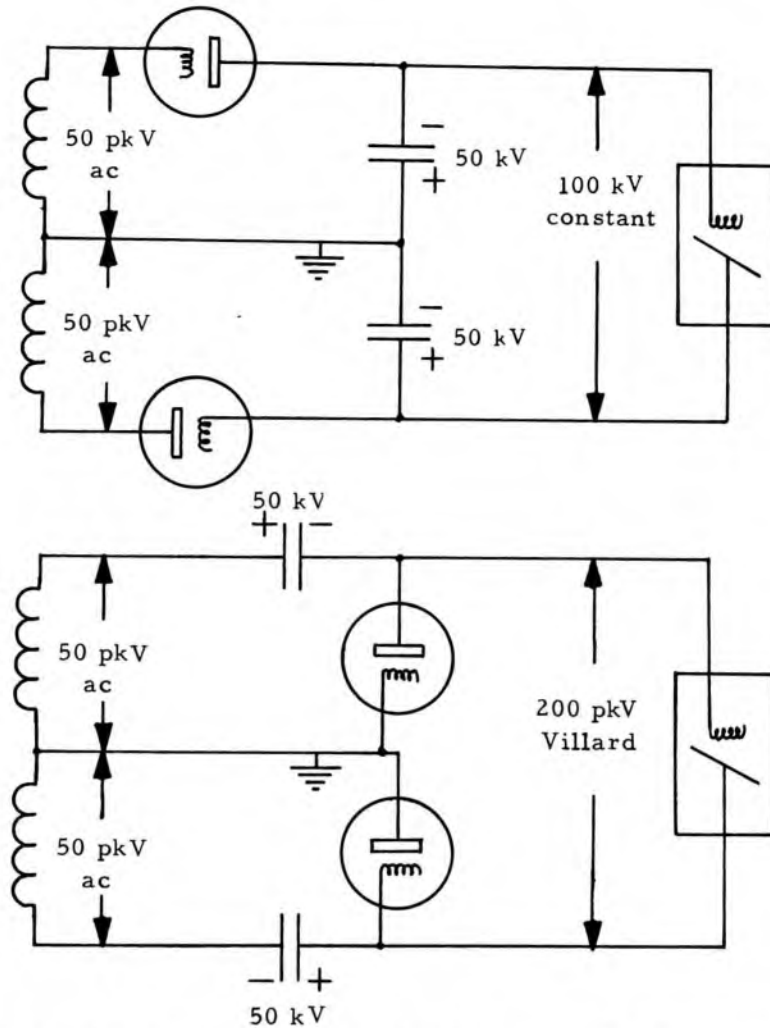


Fig. 10. Two useful applications of the basic peak detector circuit. Top. Simple constant-potential circuit uses the voltage across the *capacitors*. Bottom. Villard voltage doubler circuit uses the voltage across the *valve tubes*. Note the circuits are identical except for interchange of the valve tubes and capacitors. The top circuit produces a constant potential, of the *same* size as the transformer pkv. The bottom circuit yields a pulsating potential but with *double* the transformer pkv.

circuit followed by two peak detectors. This circuit has been employed in deep therapy to produce up to 250 kV. Note that here we have employed not only four valve tubes but also four capacitors. This is a far more expensive arrangement, especially since the final two capacitors must have double the voltage rating of the two smaller ones in the Villard part of the circuit. However, note that this circuit yields a *constant* potential of *double* the transformer peak voltage.

Overall Basic X-Ray Circuit

Figure 12 shows the basic x-ray machine circuit, with the control panel portion at the left, the shockproof cable and x-ray tube portion

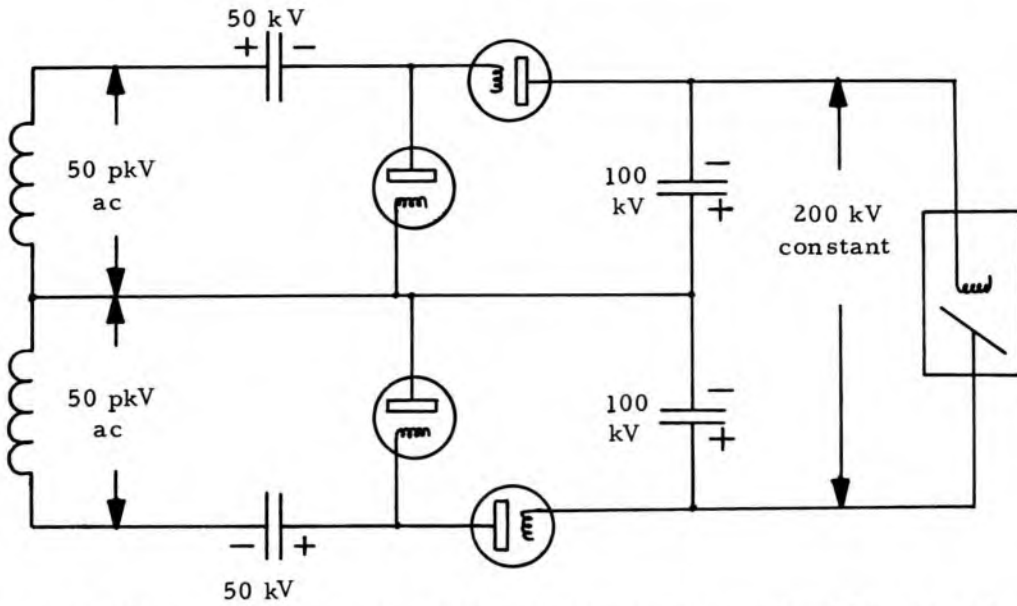


Fig. 11. Voltage doubler type constant potential circuit. This circuit uses peak detectors following a Villard circuit, so it provides both voltage doubling (100 pkV yields 200 pkV) and voltage smoothing (200 pkV yields 200 kV constant potential).

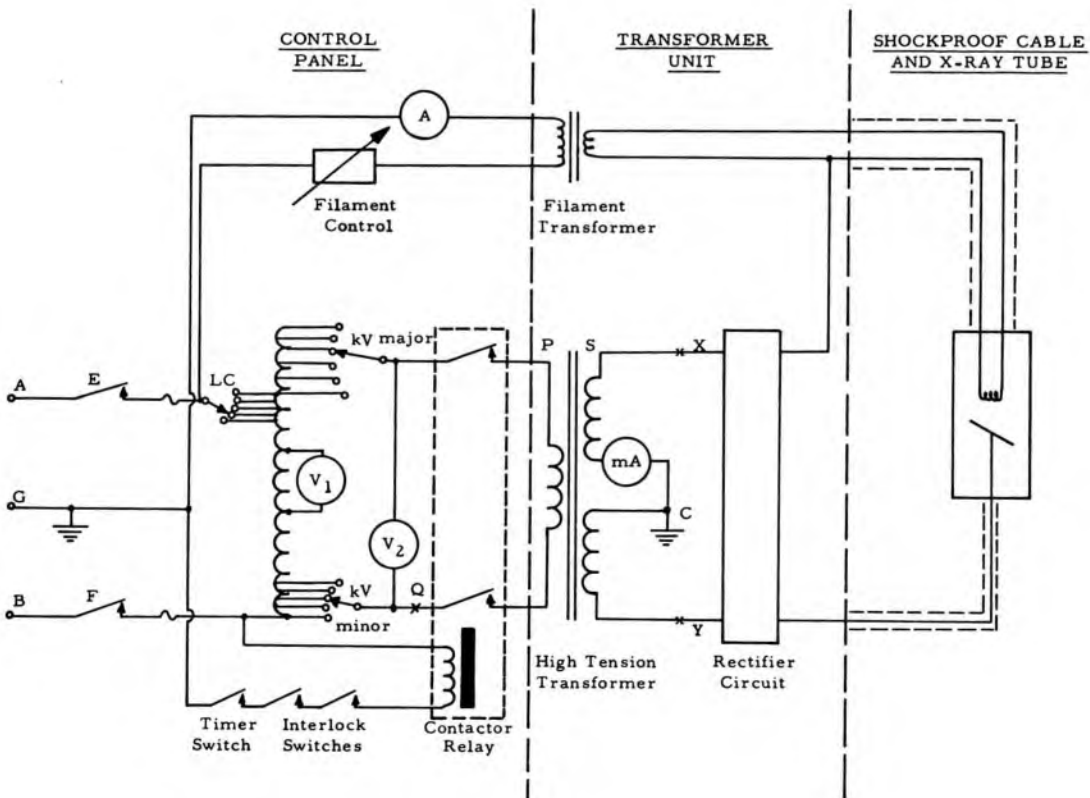


Fig. 12. The basic overall x-ray circuit.

at the right, and the high tension transformer box portion in the middle.

SWITCH CIRCUITS. The incoming 230 volt supply is customarily the usual 3-wire supply provided by the electric company. The center wire is at "ground" potential. It is connected to the center of the 230 volt winding of the electric company transformer supplying the electricity and also usually to the cold water pipes of the building. Our usual 115 volt supply for running television sets and electrical appliances is normally connected between points A and G or B and G; however, in large x-ray machines one must use the full available voltage in order to avoid drawing excessive currents. This is very similar to the situations in which a special 230 volt line is used to supply other more powerful electric units such as air conditioners and electric ranges. The main switch indicated at E and F is called a double-pole type because it contains two portions which close and open the circuit together. Beyond this are fuses or circuit breakers to protect the machine from any accidental excessive current.

The large "curled" structure to the right of the fuses is a symbolic representation of the autotransformer which is a single coil. (See page 59 for a discussion of transformer principles and designs.) (The iron core has not been indicated in this transformer for the sake of simplicity.) To the left of the autotransformer is shown the line compensation selector switch LC, and to the right are the kV major and kV minor selector switches. The selected voltage is fed to the high tension transformer primary P through the contactor switch (shown with its energizing coil below). The contactor is a double pole switch, like the main switch, but operated magnetically by the built-in coil shown. Current does not flow in this coil until several other switches are closed (the timer and safety switches referred to above). V_1 and V_2 are two kinds of voltmeters often employed in various machines for adjusting the line compensation switch. V_2 is calibrated in pkV and is referred to as the "prereading kilovoltmeter." V_1 is a blank meter except for a mark near the center; it is called the "line compensation meter" and may be used alternatively to V_2 . V_1 is employed simply to indicate when the line compensation switch is in the proper position to assure accurate pkV values.

HIGH TENSION TRANSFORMER. The high voltage is supplied to the rectifier at points X and Y. Current to the x-ray tube is measured by inserting a milliammeter (indicated mA) in the high tension transformer secondary winding circuit. This meter is of course normally located on the control panel for convenience in reading.

MA CONTROL CIRCUIT. The filament circuit usually employs either a 115 or 230 volt supply which operates the primary of the filament transformer. The filament control is used to adjust the amount of this voltage used while the ammeter (A) is sometimes used to indicate the current flowing to the filament. The filament control is usually a reactor type. This is similar to the type of control used for dimming lights in a theater. It consists of a coil into which is inserted an iron rod when one wishes to dim the lights. When the rod is all the way in, the impedance of the coil is very high and the lights (or filament) become dim; when the rod is withdrawn, the impedance drops to much lower values, and the lights (or filament) are brighter. The advantage of this type of control is that it does not get very warm and the control panel does not overheat. However, rheostats can be made to operate quite well in this position and are frequently used by some manufacturers.

HIGH TENSION RHEOSTAT CONTROL. The circuit of Figure 12 is quite general and is basically as shown for most diagnostic x-ray units. However, when voltages above about 140 pkV are employed, the high tension must be applied to the x-ray tube gradually or tube life is reduced. This can be done in a relatively few seconds, but the full high voltage cannot be applied directly without danger. To accomplish this gradual voltage buildup a special powerful rheostat is usually included at location Q. When the contactor closes, the full resistance of this rheostat is in series with the high tension transformer. Consequently, the developed pkV is limited to safe values. The rheostat resistance is then reduced to zero in stepwise fashion, finally yielding the full pkV. Both manual and motor-driven versions of this rheostat control are found in commercial deep therapy machines.

Of course this diagram can only describe the basic functions of the components in x-ray machines. A wide variety of specific designs are employed, with many different types of elaborate control circuits. As a result, the time has long since passed when radiologists or other users of x-ray equipment could profitably do their own x-ray machine service except under unusual circumstances.*

X-Ray Machine Transformers

Although we have described the general functions of transformers in the x-ray machine, the following cogent questions remain and are considered below:

* Three-phase circuits are used in some diagnostic x-ray units. They offer many potential advantages for high intensity x-ray work but are more costly.

1. Of what do transformers consist, and how do they work?
2. What determines the factor by which the voltage is transformed?
3. What are practical x-ray machine transformers like?

DESCRIPTION. A transformer consists of two coils sharing the same magnetic field (Fig. 13). A coil is an arrangement of wire in which turns are wrapped around a form as on a spool. Successive turns cannot make electrical contact with each other because the wire is carefully insulated with a complete sheath, usually consisting of a special varnish. Most commonly, turns are wound in rows which are arranged in successive layers. Current flowing in one coil produces the magnetic field.

A magnetic field is the space around a magnet, just as an electric field is the space around an electric charge. Magnets attract small pieces of certain materials like iron and various alloys but do *not* attract most other materials like copper, wood, paper, plastic, and most nonferrous materials. New magnets can be made by simply stroking or placing pieces of iron near other magnets. When this is done, it is found that some materials retain the new magnetism very well, while others lose it rapidly. In some applications, radio loudspeakers and galvanometers, for example, the ability to retain magnetism is very desirable. However, in many other applications it is desirable for the magnetism to disappear rapidly when the magnetizing agent is removed. This is the situation in various magnetic switches used in x-ray equipment, such as the x-ray contactor, and relays in general. It is evident that it is very undesirable for the x-ray contactor to remain

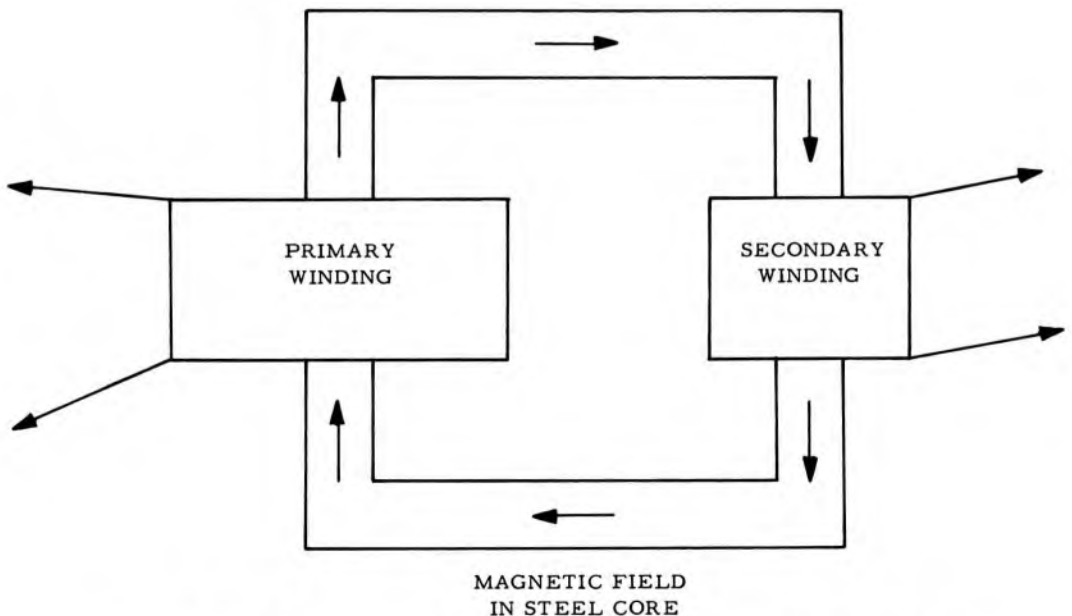


Fig. 13. The essentials of an iron core transformer.

closed after the timer signals the end of the exposure. Transformers also use magnetic materials of this controllable character; usually these are silicon alloy steels.

In a transformer the field is usually arranged in a closed loop which fits through the center of the two coils (Fig. 13). The coils may be wound on the same or separate spools, so long as the same iron ring joins them both. In some transformers, such as those used in high-frequency parts of radio and television sets, the iron is omitted.

In summary, a transformer consists of two coils connected magnetically by a closed iron loop but normally not connected to each other. An ac voltage is impressed on one of them, and a derived ac voltage is obtained across the second. The first is logically called the *primary* winding of the transformer; the second, the *secondary* winding.

OPERATION. Two aspects are involved in the operation of a transformer. The first is generation of a voltage by a changing magnetic field. Figure 14 shows a magnet placed near a coil of wire, with a galvanometer connected across the two terminals of this coil to detect any voltage produced by maneuvering the magnet. To begin we note that with the magnet at rest there is no voltage detected. Now, let the magnet approach the coil. As the magnet moves, a voltage is induced, as indicated by the galvanometer (deflection 1). Let the magnet now approach as close to the coil as possible, *inside* it. Surprisingly, once the magnet is there and at rest, the voltage becomes zero despite the fact that the magnetic field is maximum! This situation persists until we move the magnet. As the magnet is being withdrawn (and consequently as the magnetism is decreasing) a voltage is again induced, but this time *opposite* in polarity to that induced as the magnet was brought *toward* the coil (deflection 2). In general, the faster the magnet moves (i.e., the more rapidly the magnetism changes in the

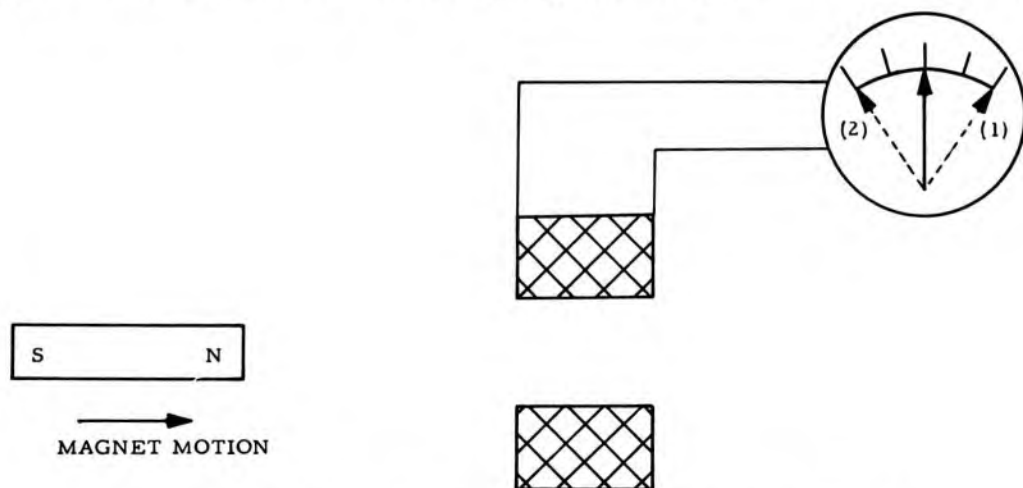


Fig. 14. Induction of electric voltage (see text).

coils) the larger the voltage produced. Also, no voltage is produced when there is no change of magnetism in the coil. It is evident that all one has to do to induce voltage in the coil is to change the magnetism associated with it.

If the strength of this magnetic field is varied in a smoothly periodic manner, the voltage induced is also smoothly periodic; in other words, it is an ac voltage. It is mechanically difficult to generate voltages by moving magnets around coils. In electric generators, voltage is therefore generated by moving coils rather than magnets.

In a transformer the variable ac magnetic field is produced by ac flowing in the *primary coil*. Figure 15 illustrates the production of a magnetic field by a coil carrying current. When switch S is closed, one observes that small pieces of iron are attracted to the coil. Remarkably, the greatest force exists where there is no copper wire at all, namely, in the center of the coil. The force can be greatly increased by inserting iron in the coil, which helps to increase the strength of its magnetic field. The field increases with both the number of turns in the coil and the current passing through it. Since the magnetic field is proportional to the current flowing through the coil, using an ac in the coil produces an alternating magnetic field in the iron. If we wish to pick up pieces of iron, a gap is required in the iron core. However, in the transformer this is not necessary, so the iron core is completely closed on itself to increase the magnitude of the magnetic field. The secondary coil is then mounted on this same iron core to share the magnetism of the primary coil (Fig. 13). Consequently, any fluctua-

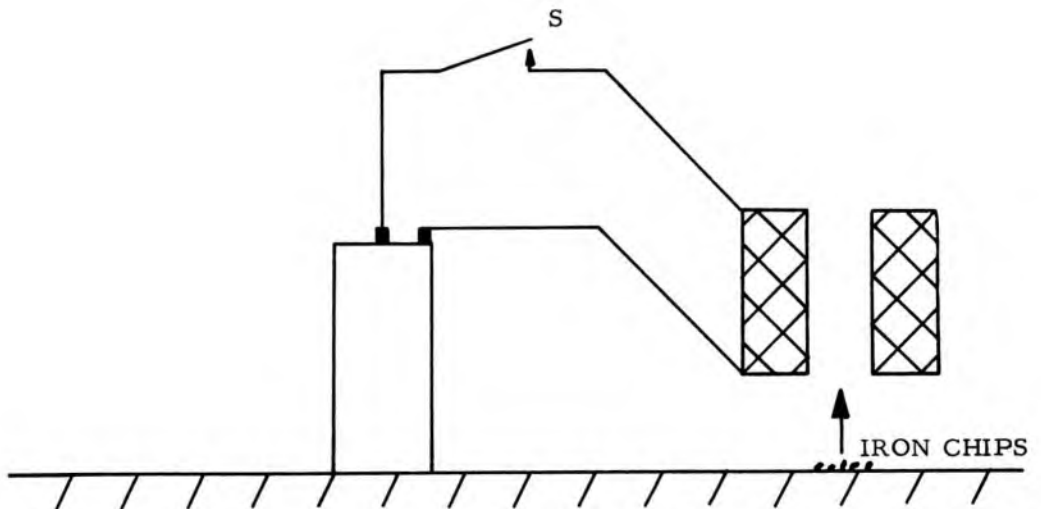
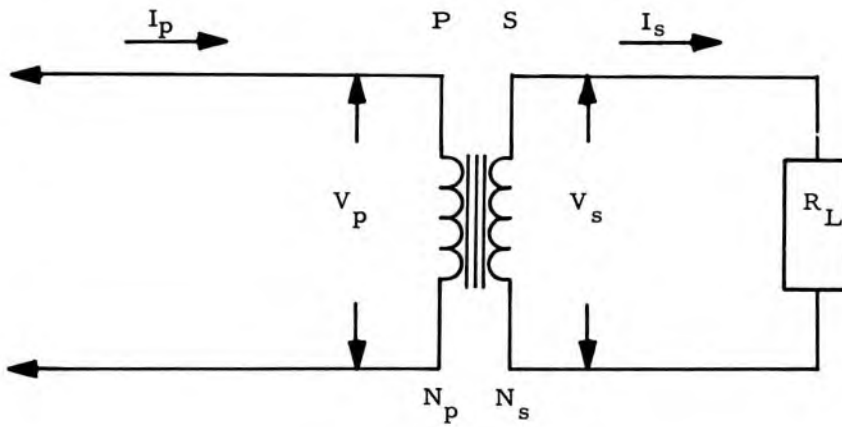


Fig. 15. Current through a coil makes it into a magnet (called an "electromagnet"). When switch S is closed, the iron chips are attracted towards the *center* of the coil. When S opens they drop again. If a soft iron bar is inserted in the center of the coil, the attraction is much greater. In relays, this attraction is used to operate switches electrically.



$$V_s / V_p = N_s / N_p$$

Fig. 16. The transformer laws. R_L represents any electrical machine or appliance. It is supplied voltage and current by the transformer secondary. The laws relate the voltage and power in the primary and secondary circuits (see text).

tion in the magnetic field produced by impressing ac voltage across the primary coil results in induction of a voltage across the secondary coil. With modern magnetic materials an excellent efficiency of such “coupling” is achieved in x-ray transformers.

TRANSFORMER LAWS. Figure 16 shows the general representation of a transformer. The primary and secondary coils are indicated by the curl-like figures to the left and the right of the vertical parallel lines. These parallel lines symbolically represent the core. So long as the magnetic coupling between the primary and the secondary is very good, as is true in all x-ray transformers, the two so-called “transformer laws” apply with reasonable accuracy.

The first law relates the primary and secondary voltages. From our description, it should be evident that the more turns we have in the secondary, the more voltage will be produced. Measurements show that the factor by which the secondary voltage is larger than the primary voltage is equal to the number of turns in the secondary divided by the number of turns in the primary. Thus, to produce a very high voltage, we should have a large number of turns in the secondary for a moderate number in the primary. For example, to *increase* the voltage by a factor of 300, we need 300 times as many turns in the secondary as in the primary winding. On the other hand, to *reduce* the voltage from, for instance, 100 to 6 volts, we need only 6/100 as many turns in the secondary as in the primary.

It is human to seek something for nothing, and high voltage generation seems superficially to provide a gift of voltage without cost. However, this is not really true. In order to increase the voltage by a

given factor, one must supply higher currents to the primary by at least an equal factor. In the above example, if we desire to produce a current in the secondary of one ampere, we require at least 300 amperes in the primary. This is because the product of the voltage and current in both the primary and secondary windings represents the energy entering and leaving the transformer each second respectively. Since the transformer can only transform but not *make* energy, the product ($V_s I_s$) can never exceed the product ($V_p I_p$) (Fig. 16). This is the transformer power law and is essentially a specific application of the law of conservation of energy.

AUTOTRANSFORMERS. Autotransformers are located in the control panel of x-ray machines and are connected to the kilovoltage selector and line compensation switches. They are essentially a single large coil wrapped around a steel yoke with a large number of wires connected to this coil at various locations. Since no high voltage appears across any of its windings, this transformer need be insulated only in the normal fashion and need not be immersed in oil. Figure 12 shows a sketch of the symbolic representation of an autotransformer and its connections to the switches mentioned above. Note that the primary in this case represents that portion of this coil between one side of the incoming line and the line compensation switch contact. The remaining parts of the coil could be considered to be the secondary winding, connected at its end to the primary winding. In order to obtain voltages greater than the supply voltage, the pkV selector switches are arranged to select as much of the total transformer winding as needed. Since the primary winding supplies part of the output voltage itself, we refer to assembly as an autotransformer.

HIGH TENSION TRANSFORMER. The high tension transformer is located in the oil-filled box with the filament transformer. As previously mentioned the high tension transformer produces the high voltage which ultimately drives electrons across the x-ray tube to produce x-rays. The primary winding carries a relatively high current, usually of the order of 10 to 100 amperes, but the voltages impressed across this winding are relatively low, of the order of 40 to 250 volts ac. The secondary winding, on the other hand, develops voltages up to more than 75,000 volts, or about 300 times as much as the primary voltage. When lower voltages are impressed across the primary, correspondingly lower voltages are produced in the secondary winding. In practice, the primary might consist of something like 300 turns of heavy wire, with roughly 300 times 300 or 90,000 turns in the secondary winding. This can be achieved only by using very fine wire for each of these turns. However, using too fine wire presents disadvantages.

The secondary of an x-ray transformer must pass currents of substantial magnitude (of the order of several tenths of amperes) in diagnostic x-ray work. As a result, excessively fine wire tends to overheat and ultimately burn out, severely limiting the available current to the x-ray tube.

The high voltage ends of the wires must be remote from each other or spark-over will occur. The windings must therefore be very carefully arranged and insulated from each other. The design and construction of this secondary winding and its isolation from the primary winding and iron core constitute a challenging problem of electrical insulation. Ceramic forms are often used for the secondary with special winding arrangements. However, even with these precautions arc-over could occur were not some insulation other than air employed within the transformer box. In most machines all electrical components are immersed in mineral oil which can safely withstand great electrical fields. In some other machines special gases (like freon) may be employed at elevated pressures, following withdrawal of air.

A word should be included about the rather often confusing concept of grounding. The higher the voltage between two given points, the greater the tendency for a spark to occur. This is because all insulating materials break down when subjected to excessive electrical stress; in solids a spark often destroys the insulation permanently. In Figure 12, suppose that 100,000 volts are developed between points X and Y of the high tension transformer. At the center, point C is 50,000 volts different in potential from either of the two ends. In practice, point C is connected to the iron core of the transformer and its case, which in turn is connected to the water pipe, radiator, and other so-called ground points in the room. The reason for this procedure is that both ends X and Y are thereby maintained only 50,000 volts away from the case, the iron core, and other items to which arc-over might normally occur.

Suppose the secondary is grounded at one end of the high tension transformer instead of at C. The other end is then not 50,000 volts from ground but 100,000 volts, and twice as much insulation is required for all items connected to it. By grounding the center of the high tension transformer secondary winding, we have thereby reduced by half the maximum voltage stress, not only on the transformer components but also on all subsequent items in the electrical circuit: the rectifiers, shockproof cables, and x-ray tube.

FILAMENT TRANSFORMER. The filament transformer converts a voltage of the order of 100 volts down to about 6 volts because it is

advantageous to employ a low voltage across the x-ray tube filament. This filament is deliberately made of relatively heavy tungsten wire coiled into a helical spiral of the order of a millimeter and a half in diameter and 1 cm long. This is a very rugged construction which renders the tube resistant to mechanical shock, an important consideration in tubes which cost up to several thousand dollars. However, this construction results in very low filament electrical resistance, necessitating correspondingly low filament voltage. The available voltage in the control circuit is considerably greater, making a substantial reduction necessary. To accomplish this, the filament transformer has relatively few turns in the secondary.

The filament transformer secondary is extremely well insulated from its primary winding. This is because the primary winding is connected to the various switches in the control panel and is therefore at approximately ground potential, give or take a hundred volts or so. The filament, on the other hand, is often about 50,000 volts with respect to ground because the high tension transformer secondary winding is grounded at the middle point. Were the filament transformer of ordinary construction, the secondary winding would arc

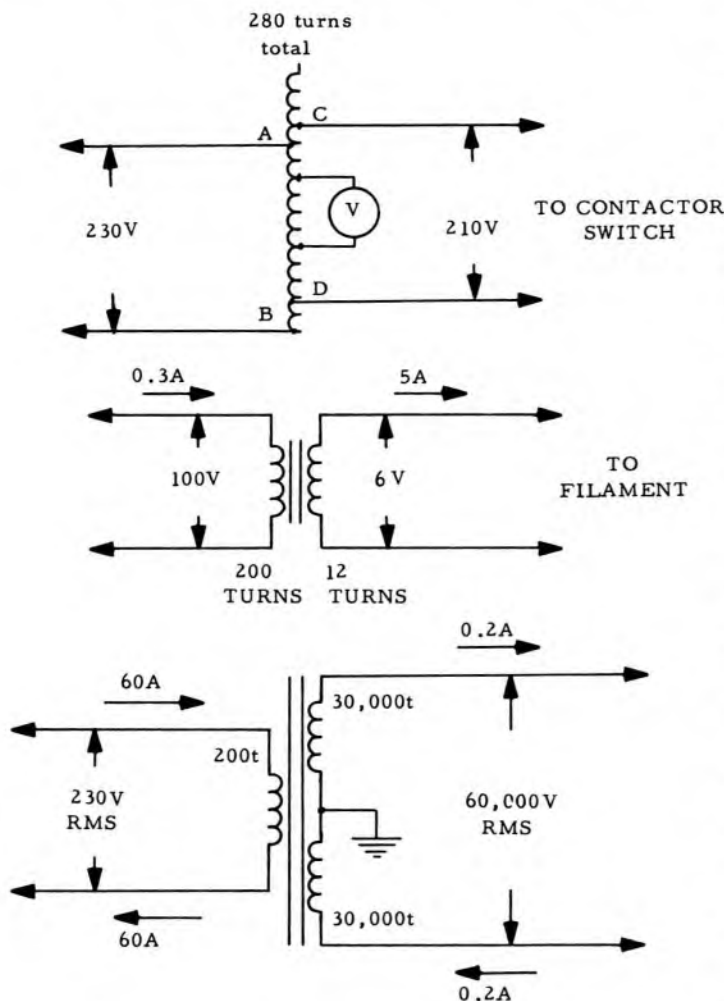


Fig. 17. Illustrations of typical values of x-ray transformer voltages, currents, and turns. Top. Autotransformer. This might operate at about 1 volt per turn. Thus, 210 V is obtained when C and D include 210 turns, provided A and B include 230 turns. The proper ratio of volts to turns is indicated by line compensation meter V. Center. Filament transformer. Note the reduction in voltage yielded by this transformer; correspondingly, a sizeable filament current is sustained by a relatively small primary current. Bottom. High tension transformer. Note the enormous increase in voltage yielded because of the great number of secondary turns, also the substantial primary winding current needed to sustain 200 mA or 0.2A tube current.

over to the primary. A very excellent insulation is therefore provided between the primary and secondary windings of this transformer, as well as between the secondary winding and the iron core.

Valve tubes employed in the rectifier unit also have filaments which are supplied by their own filament transformers. Where the secondary windings of such transformers are operated at high tension, they too must be constructed with appropriate high voltage insulation. Solid state rectifiers, of course, do not require filament transformers.

Figure 17 illustrates reasonably representative examples of both numbers of turns and voltage and current values for the three types of x-ray transformers.

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See Chapters 1 and 2 for description of machines and applications, and some references.

3

X-rays: Properties, Uses, and Production

Chapters 5 through 8 deal rather specifically with the clinical application of x-rays. To prepare for these chapters it is useful to first consider two more fundamental aspects. The first involves basic orientation: the fundamental properties and uses of x-rays. The second involves basic physics: the production and attenuation of x-rays. This chapter considers all these topics except x-ray attenuation, a major topic in its own right, which is covered in Chapter 4.

BASIC PROPERTIES (Table 1)

The properties of x-rays or any energetic particles may be considered from two points of view. The first is concerned mainly with the photons themselves, their speed, detectability, removal or dispersal by materials traversed, etc. The second considers what the photons do to objects they strike. An analogy might be the dual concern of the owner of a runaway automobile for both his car and any objects it may strike.

The Rays

X-rays travel in straight lines. This fact is very important in diagnosis because it makes possible the formation of meaningful shadow images. Also, in therapy it permits one to direct radiation accurately at a deep tumor. In common with other electromagnetic waves, x-rays travel with the velocity of light. However, x-rays are invisible and normally undetectable by other senses. Consequently, a radiation worker does not receive a sensory warning when he is irradiated by x-rays, a fact is of great importance from the point of view of radiation safety.

In common with other electromagnetic waves, x-rays are scattered

TABLE 1. BASIC PROPERTIES OF X-RAYS

A. The rays themselves:

1. **Travel in straight lines, with the velocity of light.**
2. **Are invisible.**
3. **Are scattered by materials traversed.**
 - (A) Both modified and modified scattering (see Chapter 4).
 - (B) At right angles of scatter, normally less than 1/1,000 of the incident intensity is present a meter away.
4. **Are very penetrating. Transmitted fraction depends on:**
 - (A) Thickness and compactness (mass density) of part.
 - (B) Chemical composition of part.
 - (C) Kilovoltage and filtration of x-rays.

B. What x-rays do:

1. **Ionize atoms—basic action.**
 - (A) Primary ionization—directly by the photons.
 - (B) Secondary ionization—indirectly by fast electrons released during primary ionization.
2. **Ionize gases—basis for accurate measurement.**
3. **Produce solid state effects.**
 - (A) Fluorescence—basis for fluoroscopy, most radiography, and scintillation detection systems.
 - (B) Other effects are used in x-ray dosimetry—photoconduction, thermoluminescence, and photoluminescence (see Chapter 8).
4. **Produce chemical effects.**
 - (A) Photographic film activation.
 - (B) Fricke dosimeter (Chapter 8).
 - (C) Biologic action (Chapter 5).

in all directions when they impinge on objects. However, unlike light and radio waves, x-rays are changed in wavelength when scattered (Chap. 4). It is useful to note that x-ray dosage (Chap. 5) from scatter to a person, one meter from an irradiated patient at right angles to the beam, is generally less than 1/1,000 that received by the patient himself. While this seems like a relatively small amount, repeated irradiation by scatter is a potentially serious occupational hazard.

Scatter adversely affects both x-ray diagnosis and therapy. In diagnosis, it blurs images by adding a relatively uniform exposure to the entire film (Fig. 1). This is similar in effect to light fog on an ordinary photographic film and is often referred to as "scatter fog." In radiation therapy scatter irradiates areas not in the direct treatment field, so that adjacent healthy tissues are inevitably irradiated along with the tumor.

The penetrating character of x-rays was one of the first attributes noted by early investigators. Of equal importance is the fact that the degree of penetration depends on the type and thickness of the object

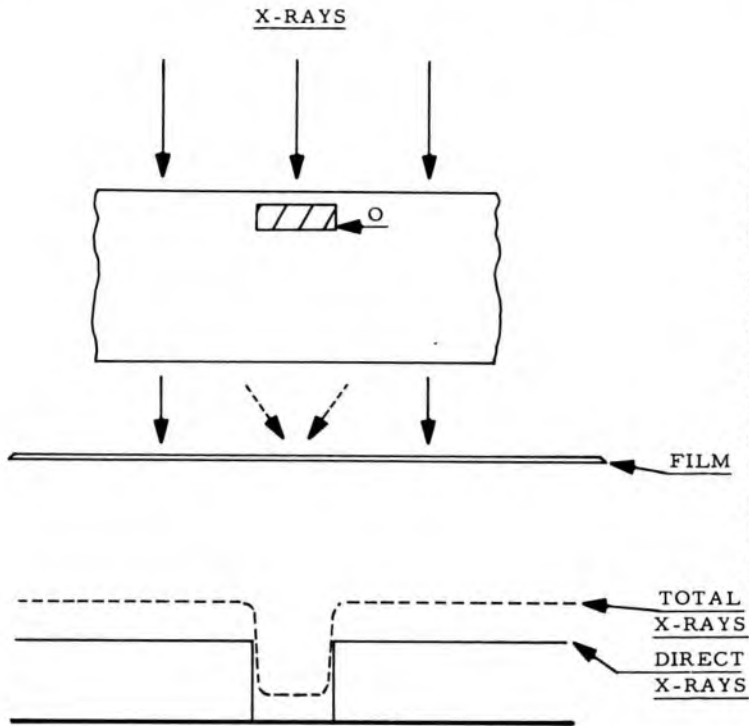


Fig. 1. How scatter reduces image contrast. Top. Exposure of film through subject containing opaque object O. Bottom. Distribution of radiation reaching film. Solid lines represent directly transmitted x-rays; dashed lines represent total to film. Note substantial radiation reaches the area behind O which should be in total shadow.

traversed by x-rays. This *differential* penetration is the basis for x-ray diagnosis, resulting in the shadow images so important in medicine and material testing. The penetration of an x-ray beam depends on the following:

1. *Thickness of the part.* For example, a person's forearm transmits x-rays to a greater extent than his adjacent thigh.
2. *Mass density of the part* (g/cm^3). For example, lung tissue is penetrated by x-rays more than the adjacent hilum, despite the fact that tissues in both are of similar chemical composition. Lung tissue is less dense because it contains air which transmits x-rays readily.
3. *Chemical composition.* The x-ray transmission through an object of given thickness and density depends on the atomic number Z of the elements present and their relative amounts. It is unimportant what the actual specific chemical form of the combination is. For example, it is noteworthy that a 50 percent solution of alcohol transmits x-rays identically with a certain noxious mixture of cyclopentene and water (Fig. 2).

The dependence of x-ray penetration on the nature (Z) of the material traversed is of great clinical significance. Thus, soft tissue is more transparent to x-rays than adjacent dense bone of the same total mass, and fatty tissue is more transparent than glandular or muscle tissue. The dependence on chemical composition is most pronounced at low photon energies (Chap. 4).

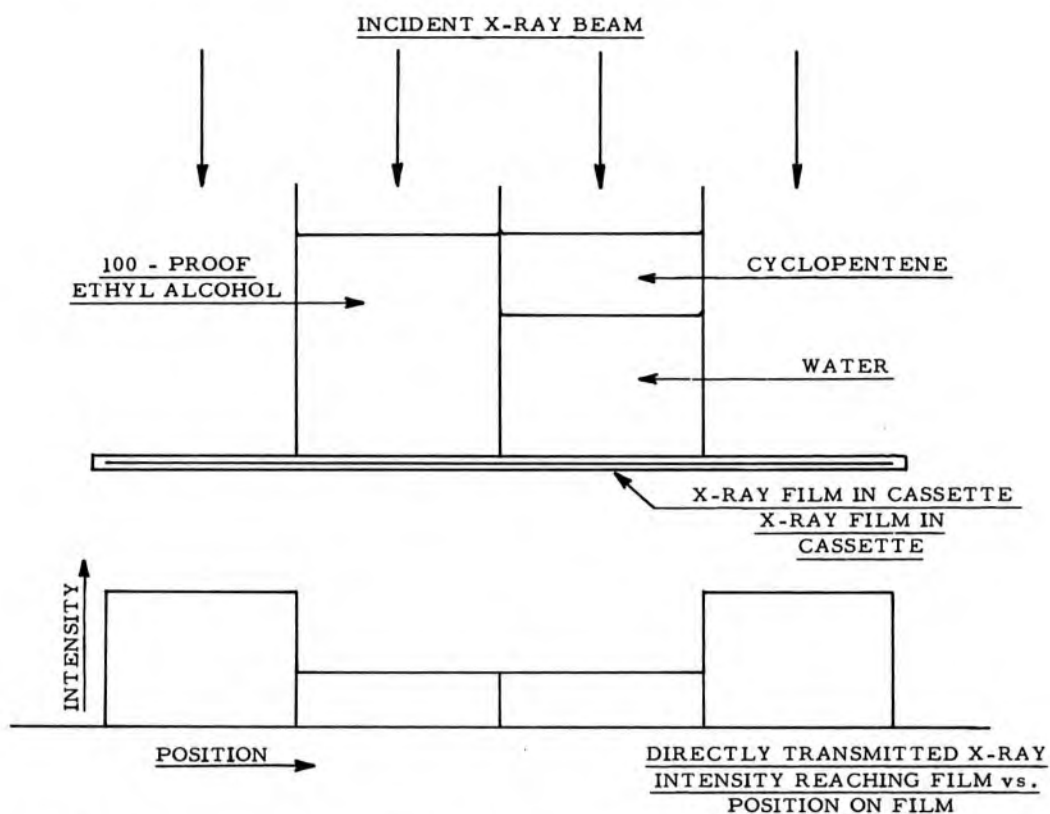


Fig. 2. X-ray transmission depends on the *kind* and *relative amounts* of atoms traversed but not of their specific chemical form. Top. The left container has the same amount by weight of carbon, hydrogen, and oxygen as the right one, but important aesthetic differences exist between the two liquids. Bottom. The radiograph shows identical transmission of the two liquids. Roentgenograms are evidently unsuitable for evaluating the form of chemical combinations.

Effects of X-rays

X-rays produce their effects by ionizing atoms. When it interacts with an atom, a photon gives up some or all of its energy to one of the electrons, which escapes from the atom's influence. The electron is generally released with substantial energy of motion. By means of multiple collisions, the electron gives up this energy to other atoms in its path, ionizing and exciting many of them.

In this book, we denote ionization produced directly by a photon as *primary* and that produced by the released electrons as *secondary* x-ray ionization. It is evident that most x-ray ionization is of the secondary type, since a single primary electron may ionize a great many atoms.

The ionization of gases provides the basis for accurate measurement of x-rays by ionization chambers, employed for dosage measurements in both x-ray therapy and diagnosis. X-rays also activate photo-

graphic films, which can be developed as in ordinary photography. Perhaps even more important in diagnostic radiology is fluorescence, which is the production of visible light when ionizing radiations strike certain crystals. This light is actually visible characteristic radiation, arising in outer orbits of ionized atoms (Chap. 1). Fluorescence is the basis for both fluoroscopy and almost all clinical radiography.

X-rays produce the photographic effect by chemical action. Other chemical effects have also been employed for measuring x-ray dosage. One of the most useful is the oxidation of ferrous to ferric ion, in the Fricke dosimeter (Chap. 6).

Perhaps the most important chemical effects of x-rays and other ionizing radiation are those produced in living materials. Since the ionizing actions of x-ray interactions occur on an atomic level, resulting chemical changes may be produced within cellular constituents and are hence extremely effective biologically. The net result is to alter some essential cellular constituents and produce new and potentially toxic ones.

The biological effects of ionizing radiation do *not* result from heat production. To illustrate, consider an unfortunate man subjected to a uniform dose of 600 rads (for discussion of unit of radiation dosage, see Chap. 5) to his entire body in a disaster situation. He will probably die within 30 days, regardless of what heroic measures may be taken. Yet, his body has received a total radiant energy no greater than that required to raise his temperature $.0015^{\circ}$ C! Obviously, these approximately 100 calories* total ionization energy were far more destructive than 100 calories of heat would have been. (One receives more heat from a cup of coffee.)

The reason lies in the site of application of energy. In the case of heat, energy tends to become uniformly distributed in tissue, and normally cells are thermally well-buffered by surrounding fluids. However, ionization represents a very great local concentration of energy. This is easier to appreciate if one considers a mechanical analogy of a man resting on the floor with a heavy book on his chest. The book presses down with reasonable force and does a certain amount of work in displacing his anterior chest wall but inflicts no injury. An equal amount of work is performed by a rapier placed against the man's chest and gently pushed into his heart, but with a dramatically greater biological effect. The different result is due to the specific site of application of energy deposited. Similarly it is the specific site of application of ionizing radiation within the vulnerable parts of cells which accounts for its remarkable biological effectiveness.

* If 1 rad is 100 ergs/gram of energy in tissue, 600 rads is 60,000 ergs/g, or $60,000/10,000,000$ joules/g = $.006$ joules/g. Since 1 calorie is 4.185 j/g, this is $.0015$ calories/g. For a 70,000 g man, this yields about 100 calories total.

CLINICAL APPLICATIONS

Because x-rays are transmitted differently by different parts of the body, an x-ray beam acquires useful information when it traverses a patient. However, this information must be made visible to help in diagnosis. Two basic x-ray detection methods are employed to accomplish this objective: fluoroscopy and radiography. The former uses fluorescent materials to derive a visual image; the second employs photographic film to produce a permanent record, almost always a negative or transparency which is viewed on a bright background (using a "viewbox"). Special modifications of these basic systems are also employed.

In this section we describe the basic setup, advantages and limitations of fluoroscopy, radiography, and other diagnostic procedures. In addition, brief comments are given on special techniques used to obtain maximum information from diagnostic studies. This and other aspects of x-ray diagnosis are covered more fully in Chapter 8. Finally, a short introduction is given to the philosophy of the therapeutic application of x-rays.

Fluoroscopy

Figure 3 shows basic components of a fluoroscopy unit. X-rays are directed through the table panel toward the patient and a fluoroscopic screen assembly beyond him. The viewer observes the light image produced by transmitted x-rays when they strike a fluorescent screen in the fluoroscopic assembly. The size of the x-ray beam, usually referred to as the "field," is adjusted by the use of lead shutters located beneath the table. The x-rays produce light upon striking the fluorescent screen but cannot irradiate the doctor beyond the screen assembly because they do not significantly penetrate the lead glass shield. The fluorescence is proportional to the x-ray intensity, so the visible image corresponds to the distribution of x-ray intensities in the beam emerging from the patient.

The major advantage of fluoroscopy is the almost instantaneous picture presented of x-ray transmission by the patient's anatomy. Consequently, internal organs can be observed in motion and the position of the patient adjusted for the right view at precisely the right time to secure the most useful information. Unfortunately, there are accompanying disadvantages. Since the image is ephemeral, it cannot be directly compared with that observed at previous examinations.

Two other disadvantages limit the usefulness of simple fluoroscopy. First, both detail and contrast are inherently much poorer than

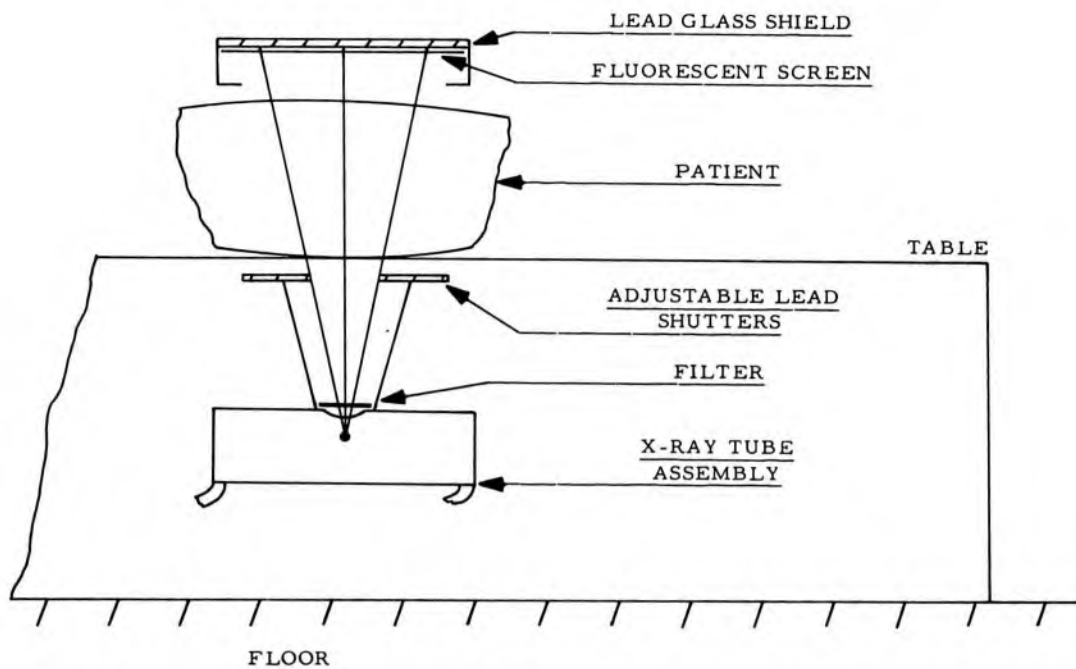


Fig. 3. The setup in fluoroscopy. Starting from tube and working upward: The tube assembly moves with the screen assembly, so the x-ray beam is always stopped by the shield. The tube assembly can move both across and parallel to the length of the table. A filter of 2 or 3 mm Al is required to protect patient's skin. Adjustable lead shutters limit the rectangular x-ray field size. The fluorescent screen glows when struck by x-rays. The lead glass shield absorbs x-rays but transmits light from the screen to the viewer beyond the screen. The fluorescent screen assembly often accommodates fixed or moveable grids for scatter control and cassettes for spot or flash films (not shown).

that achieved using x-ray films (radiography). Second, the patient dose is considerably higher in fluoroscopy than in most radiographic procedures. To illustrate, the skin of a patient may typically receive about 5 rads during each minute of fluoroscopy. By contrast, the dosage delivered to a patient during a chest radiograph is only about $1/25$ of a rad; the patient receives more dosage to his skin from one second of fluoroscopy than from a chest radiograph!

Radiography

In radiography (Fig. 4) x-rays traversing the patient expose an x-ray film which is then developed yielding a permanent record of the x-ray image. As mentioned above, the visible detail is much greater than that obtained in fluoroscopy. However, the image corresponds to only one short interval of time during which the x-rays traverse the patient. Unless the particular time is properly chosen, one may err in an important diagnosis by missing an event occurring before or after the exposure.

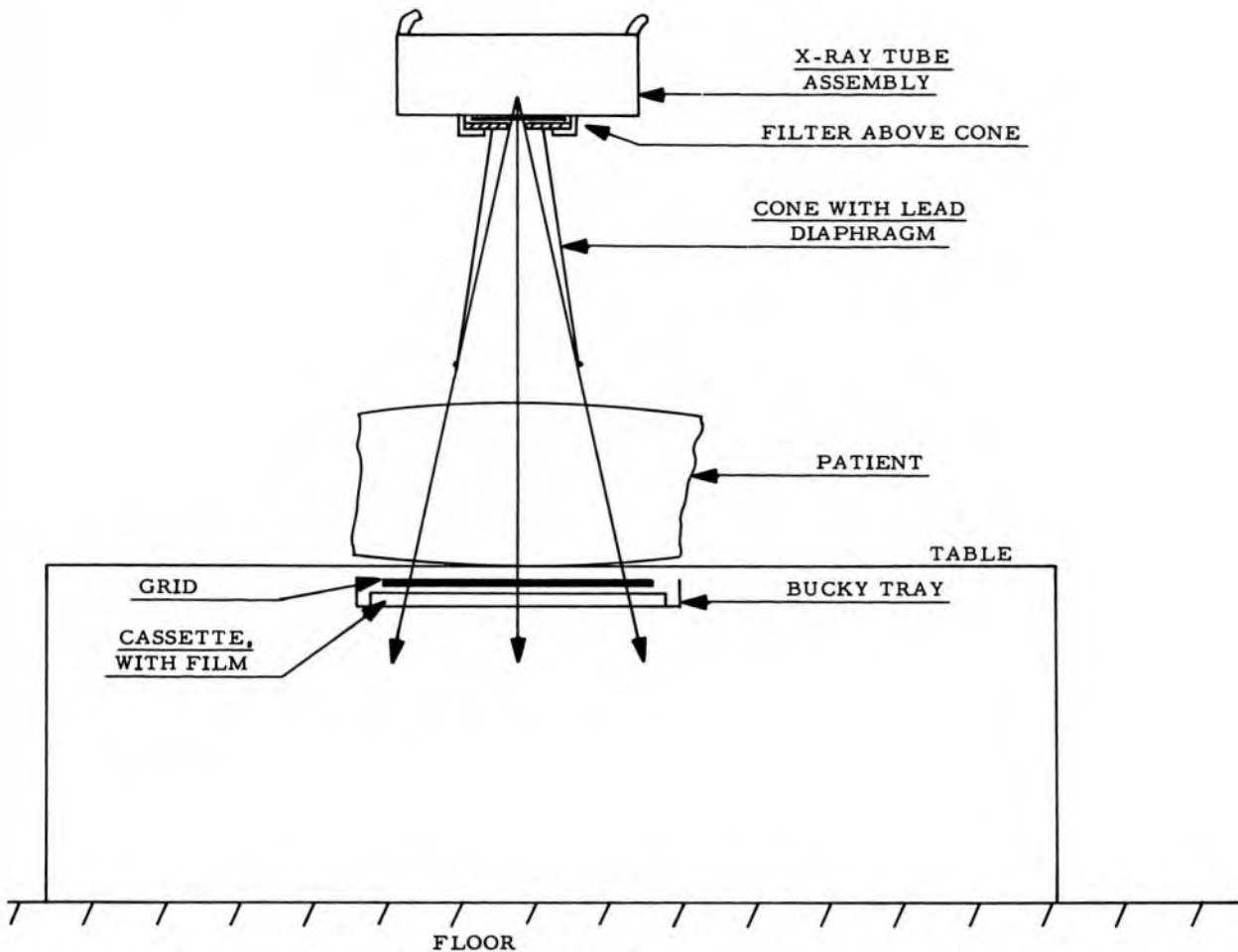


Fig. 4. The setup in radiography. Starting from tube, working downward: The tube assembly supports the filter of 2 or 3 mm Al and the cone. Cones have lead diaphragm at top to determine the field and a sheet metal assembly below to *indicate* the beam field. Bucky tray holds grid and cassette and can move lengthwise along table; cassettes are inserted from side. Cassette contains film between intensifying screens, when screen techniques are used.

To partially overcome this limitation, one can combine fluoroscopy and radiography by inserting a cassette containing an x-ray film ahead of the fluoroscopic screen to record an important finding observed during fluoroscopy (Fig. 3). Such a radiograph obtained during fluoroscopy is usually referred to as a "spot film" or "flash film" (depending on the technique used) and can be very helpful to the diagnosis.

Technical Aspects

SAFETY. The x-ray tube assembly not only mechanically supports and shields the x-ray tube but also has two other functions. First, it is "rayproof," meaning it contains sufficient lead to absorb x-rays emerging in directions other than the useful one. (The shielding has an

opening for x-rays traveling in the desired direction.) Second, modern x-ray housings are shockproof, so they may be touched during exposure without shock hazard. However, it should be stressed that many older x-ray tubes are neither shockproof nor rayproof and should be replaced as soon as practical with modern tubes.

A filter is required in both fluoroscopy and radiography. This is normally a piece of two or three millimeter thick aluminum placed near the tube opening. For reasons discussed in Chapter 4, a filter selectively removes photons which are superficially absorbed and contribute unduly to skin injury, but little to the roentgen image.

SCATTER AND GRIDS. Scattered x-rays tend to produce a uniform fog or blurring of the image. Two basic ways exist to control this undesirable effect. The first is to reduce the total amount of scatter by reducing the volume of the patient's body irradiated. Obviously, the fewer incident x-ray photons, the fewer that can be scattered. In radiographic work this is often accomplished by means of fixed lead diaphragms attached to so-called "cones" (Fig. 4). In fluoroscopy as well as in some radiographic work, more versatile adjustable lead diaphragms are employed.

In many situations, the size of the object being radiographed is too large to permit very much field reduction (for example, in examinations of the chest and pelvis). Under these and some other circumstances, one must try to reduce scattered radiation reaching the detector without reducing the total amount produced by the patient. This can be accomplished by using so-called "grids" placed directly in front of the detector.

A grid contains a large number of thin, lead strips arranged parallel to the x-ray beam. The strips are separated by paper, plastic, or aluminum strips which are relatively transparent to x-rays. Figure 5 shows the action of a grid in removing scatter. Note that direct radiation shown by the solid lines traverses the grid substantially unabsorbed, except by the thin lead slats. On the other hand, scattered radiation generally has been altered in direction so that it strikes the slats of the grid obliquely and is absorbed. As a result, the grid removes a significant part of the scattered radiation before it can strike the x-ray film or fluorescent screen.

In some situations the grid is simply placed ahead of the detector and left there. This is called, for obvious reasons, a "fixed" type of grid. When the grid is fixed, its lead slats are bound to cast shadows. These may be noticeable on the film and are called "grid lines." Where they are objectionable, the grid is moved perpendicularly to the lead slats during the exposure to blur images of the lead slats. When properly carried out, so-called "bucky" grid films show no noticeable grid lines.

New "fine-line" grids with more than 100 lines/inch cast such fine shadows that their use is often considered acceptable without moving the grid.

Note in Figure 5 that an ordinary grid can intercept only those x-rays that strike obliquely. Such a grid therefore transmits scattered radiation traveling parallel to the face of the lead strips (ray 5). As a result, a significant amount of scatter slips through the grid. To correct this, one can employ two grids placed one above the other, with the lead strips at right angles. By this means, much more complete clean-up of scatter is obtained. The use of such so-called "crossed grids"

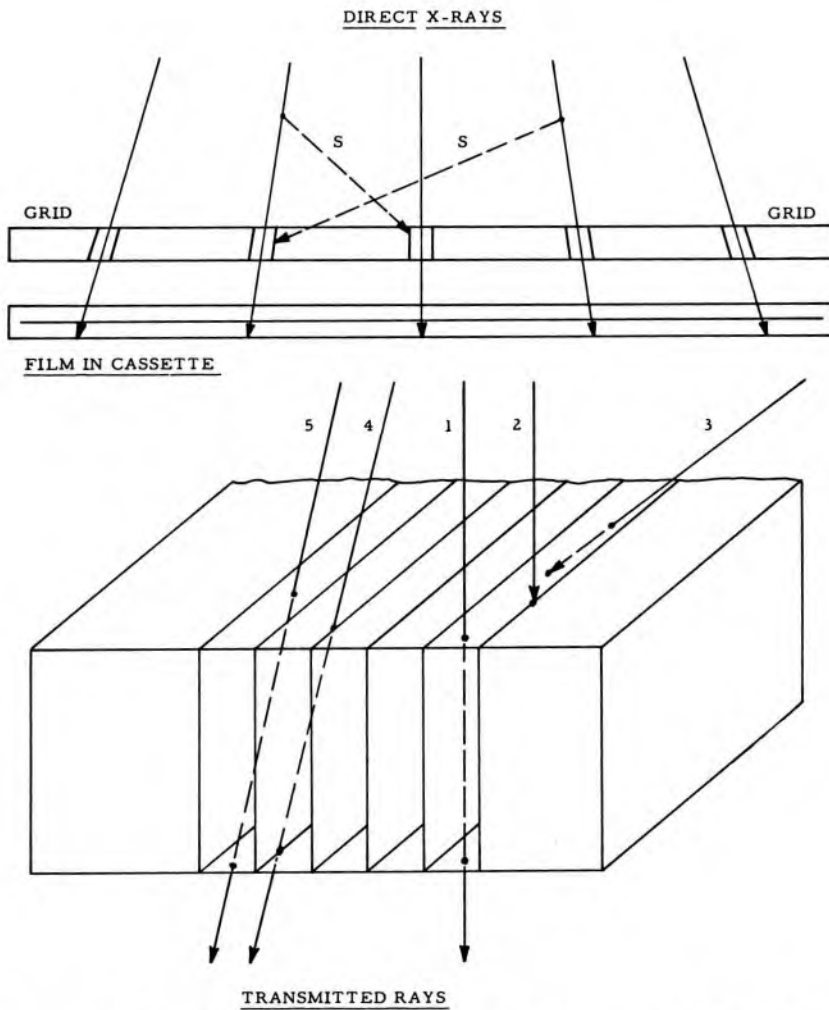


Fig. 5. How a grid reduces scatter to the detector. Top. The thin lead slats in the grid (only a very few shown) remove *oblique* radiation, which is scatter (S). Since direct rays are parallel to the slats they are transmitted except for the slats struck edgewise. Bottom. A more complete picture of what happens—five rays considered. 1. Direct ray between slats—transmitted. 2. Occasional direct ray hits slat edgewise—removed. 3. Oblique scatter ray hits slat—removed. 4. Very slightly oblique ray—transmitted. 5. Oblique ray in plane parallel to slats—transmitted. (Stopped only by "crossed grid.")

has the disadvantage that they must be positioned very carefully (both in centering and angulation) to avoid grid lines. For this reason, their use is impractical in most routine radiography.

FLUORESCENT SCREENS AND CASSETTES. Fluorescent screens are commonly made of a cardboard or plastic sheet to which fluorescent crystals are bonded. These crystals are usually zinc sulphide with a tiny amount of cadmium sulphide added as "activator" to increase the light production. The transparent glass shield above the fluorescent screen contains considerable lead oxide which very effectively removes x-rays. However, since this oxide has been chemically combined with silica to yield a glass-like material, the shield is transparent to light produced by the fluorescent screen.

Although some x-ray films are designed for direct irradiation by x-rays, their medical use is generally confined to the body extremities. This is because direct film exposure is slower than the so-called "screen technique." As a result, required exposure times are excessive for most studies.

More generally, an indirect method of exposing the x-ray film is employed which reduces the required exposures at least ten times. Two fluorescent screens are placed in intimate contact with a light-sensitive type x-ray film ("screen film"). The film is made with thin emulsions on both sides, to reduce the required x-ray exposure. Figure 6 shows an "intensifying screen" type of cassette in cross section with a film in place. Consider what happens when a narrow beam of x-rays impinges upon this assembly. It traverses the Bakelite cover (A) rather easily, penetrates the proximal screen (B), then the film (C), and finally the distal screen (D). It may or may not penetrate the back metal cover (F), depending on the photon energies involved and the nature of the cover. The x-rays act directly on the film emulsions where they strike them on both sides and produce a slight latent image. However, the major activation of the emulsions results from light produced by the two adjacent intensifying screens, which glow as a result of the passage of x-rays through them.

In practice, all but 3 percent or less of the final film image is due to light from the intensifying screens. As a result, intensifying screen cassettes are generally of the order of 10 to 40 times faster than even the fastest direct-acting x-ray film. However, the resulting images are not so fine in detail as those obtainable with the slower direct action films. The reasons are discussed in Chapter 8.

Other X-ray Detecting Systems

In the screen cassette, intensifying screens are placed in direct contact with the film. This is the most efficient optical contact possible

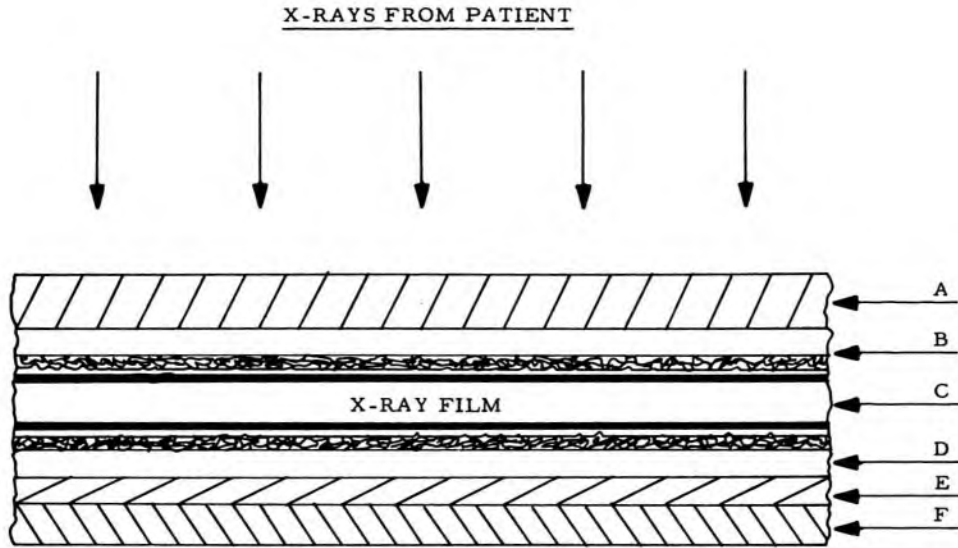


Fig. 6. Cross section of an intensifying screen cassette (not drawn to scale). A. Front Bakelite cover. B. Front intensifying screen; phosphor crystals in plastic binder, mounted on plastic. C. Film, emulsion on both sides (thick black lines). D. Rear intensifying screen, similar to front. E. Felt padding to help assure good screen-film contact. F. Metal support and exit shield for cassette.

and yields the fastest possible exposure. However, if one is willing to sacrifice some light, he may simply use a suitable mirror or lens system between the intensifying screen and the film. Figure 7 shows the principle employed in a photofluorographic machine. This essentially employs a fluorescent screen and camera combination of special design. The screen used emits light of a color to which the film is most responsive. Usually a special optical system is employed to increase the light-gathering ability of the camera. Such photoradiographic machines have been very widely used in the familiar chest survey studies carried out for early detection of tuberculosis, because the cost is much less than that of full cassette chest radiography. The photoroentgen film is only 70mm wide in newer machines and as small as 35mm in older machines, so the information yield is somewhat less than that obtained from usual technique.

Image intensifier systems are widely employed in radiology to brighten fluoroscopic images. Sometimes the radiologist views the image directly in a reflecting mirror, but more frequently he uses a television display. He can then perform fluoroscopy in daylight, with greater visual acuity than is possible in the darkened room required for ordinary fluoroscopy. Intensifier systems may also be used with movie cameras (cineradiography) and other recording means to obtain a permanent record. Thus, image intensifiers make it possible to combine advantages of fluoroscopy and radiography in the same study.

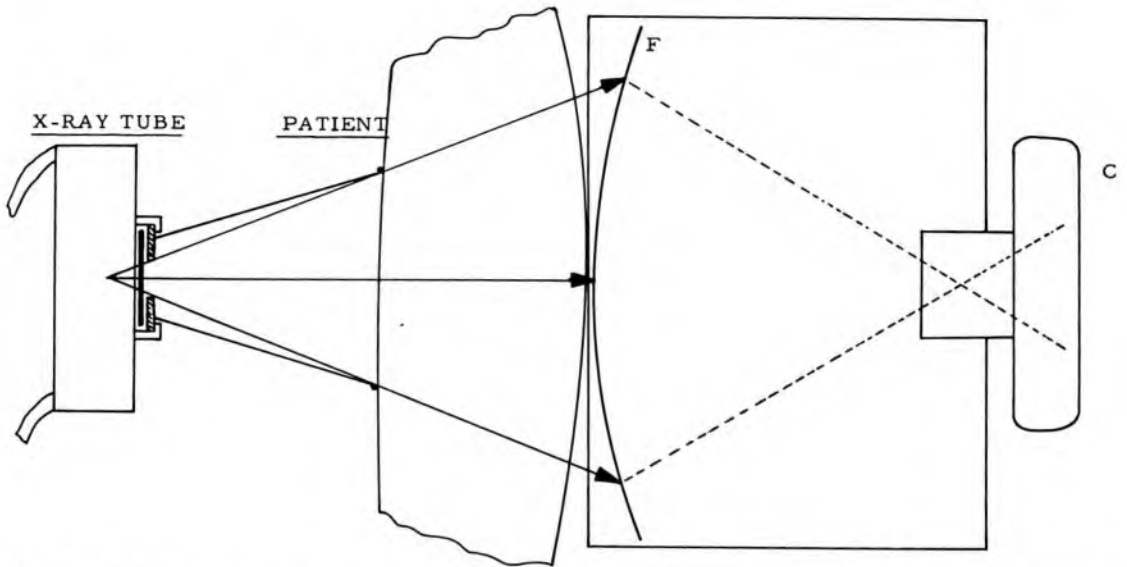


Fig. 7. Photoroentgen setup (not drawn to scale). X-rays (arrows) penetrate both patient and the front of a light-tight box, producing a fluorescent light image on the screen F. Light from this image (dashed lines) exposes 70 mm roll film in camera C. Lead shielding is provided for both scatter control and radiation protection (not shown). In practice, modern special optical systems are used which gather light more effectively than the simple camera shown.

Contrast

In both fluoroscopy and radiography, there must be a difference in the transmitted x-ray intensity of adjacent body parts for them to be distinguished. The fractional difference between these intensities is referred to as “contrast.” (See Chap. 8.) Without contrast the final image would simply be a uniformly bright blank. The contrast present in the x-ray beam leaving the patient or object is referred to as “subject contrast.” As previously mentioned, this is dependent upon both the part roentgenographed and the photon energies employed.

Differences in x-ray transmission by different tissues of the human body produce natural x-ray contrast. Table 2 lists one group of body components of generally *similar* transmission and another smaller group with *atypical* transmission. The differences among these body components provide the physical basis for seeing anatomy in ordinary x-ray studies. Unfortunately, many important situations exist where natural subject contrast is insufficient. For example, one cannot normally differentiate among the following: tumor, fibrous and glandular tissues, tissue fluids, and blood vessels.

The most important natural contrast materials are air, fat, and bone mineral. Natural contrast is enhanced by low kV techniques, but they are normally limited in application to the breast and body extremities.

Special techniques have been developed to insert opaque materials into various body cavities and vessels. These are called contrast techniques. Barium sulphate and appropriate iodinated oils or organic compounds are commonly used agents (Table 2). On occasion air or other gases also serve as contrast agents.

Special Exposure Procedures

In the approximately seventy years since the discovery of x-rays, special procedures have been devised to glean more information from x-ray studies. Three of the most notable will now be mentioned. The first is the technique of multiple view of the same area of interest.

TABLE 2. NATURAL AND ARTIFICIAL CONTRAST

A. Natural Contrast

1. Body parts of similar absorption

- (A) Body fluids: water, blood, lymph, cerebrospinal fluid; cyst contents; urine, etc.
- (B) Predominantly non-fat, non-calcific solid tissues:
 1. muscle, skin, mucosa, serosa, blood vessels.
 2. glandular tissue, visceral organs in general.
 3. nervous tissue generally, where fat content is small.
 4. most tumor tissue.

2. Body parts of atypical absorption

- (A) Atypically high—bone, calcific deposits
- (B) Atypically low—tissues with fatty inclusions: breast, kidney fat pads, etc.
- (C) Very low absorption—tissues enclosing air: pharynx, trachea, bronchi, lung; gas bubbles in G-I tract, etc.

B. Artificial Contrast—materials and use

1. **Barium sulphate**—as aqueous suspension, administered for upper and lower G-I tract examination
2. **Organic compounds with up to 50 percent iodine**
 - (A) Metabolic concentration following infection—gall bladder; intravenous urogram study of kidneys, ureters, and bladder.
 - (B) Direct insertion into organ—retrograd pyelogram, sialogram; salpinogram; bronchogram; following fluid removal, myelogram.
 - (C) Blood circulation studies—selective opacification of arterial and venous systems of most major body organs, using catheters.
 - (D) Lymphography.
3. **Air or gas**
 - (A) After fluid removal—brain ventricles.
 - (B) Under pressure, for distension and mucosal pattern study—colon, stomach, etc.

Very often manipulation of the patient permits the obtaining of a particular view showing details not visible from other directions.

The second technique is stereoscopy. This involves immobilizing the patient and taking two successive roentgenograms: the one is exposed from a slightly different angle than the other, corresponding to the interpupillary viewing angle. When the resulting films are simultaneously viewed in an appropriate manner, the illusion of depth is obtained. This technique has been extensively employed in studies of the skull as well as of other anatomic areas.

The third technique is body section radiography, or "tomography." While several tomographic systems are used, all produce a radiograph essentially of a single narrow slice of the patient, with greatly reduced contrast of body parts on either side. This is done by appropriately moving the film and x-ray tube about a fixed point, or "fulcrum," in some versions, or rotating the patient in others, during the x-ray exposure. The image of all parts except those in the desired plane or axis are blurred by the motion and hence appear indistinct on the film. A relatively clear image results in the so-called "focal plane," but images of parts on either side are blurred out.

These topics will be considered in more detail in Chapter 8.

Diagnostic Skill

X-ray studies require maximum skill of all those carrying out the examination. A technically inadequate radiograph imposes inherent information limits, so that a skillful x-ray technologist is required.

Both interpretation of all radiographs and carrying out of fluoroscopy require a highly trained physician. A certified diagnostic radiologist is a licensed physician who has pursued a residency of three years and a year's preceptorship, followed by successful completion of a comprehensive specialty board examination. His training involves not only a thorough study of the physical and technical aspects of ionizing radiation but also unique medical disciplines including living roentgen anatomy and pathology, which differ substantially from those of the cadaver.

Therapeutic Use

Ionization is by far the most effective therapeutic agent currently available for treatment of malignancy, and the hoped-for simple universal cancer cure is still apparently not in sight.

The objective of radiation therapy is primarily to destroy malignant tumors, although a few benign conditions are still treated by irradiation. However, this is only half the story. To be of real benefit,

the destruction of a malignant tumor must be accomplished without excessively compromising the comfort and useful activity of the patient. Achieving both these objectives requires considerable creative planning by the radiotherapist.

It should be stressed that the radiation reaction is highly complex, with many very subtle features. Much research is in progress regarding the biological effects of radiation; these may yield clues to more effective radiation therapy in the future.

BASIC APPROACHES. Figure 8 (Left) shows a single supervoltage therapy portal directed toward a tumor deep within a patient. Generally, such a single portal always delivers a smaller dose to the tumor than to the intervening healthy tissue because the tumor is more remote and shielded by tissue. As a result, deep tumors cannot be treated to high dosage levels by means of a single portal.

To get around this difficulty, the beam is aimed at the tumor from several directions [Fig. 8 (Right)]. The individual beams all contribute to the dose at the tumor location because they converge there, but those to surrounding tissues, like skin, do not coincide so their dose totals are lower. By this means it is possible to deliver high doses to deep-seated tumors while preserving the surrounding tissues adequately for reasonable patient function and comfort.

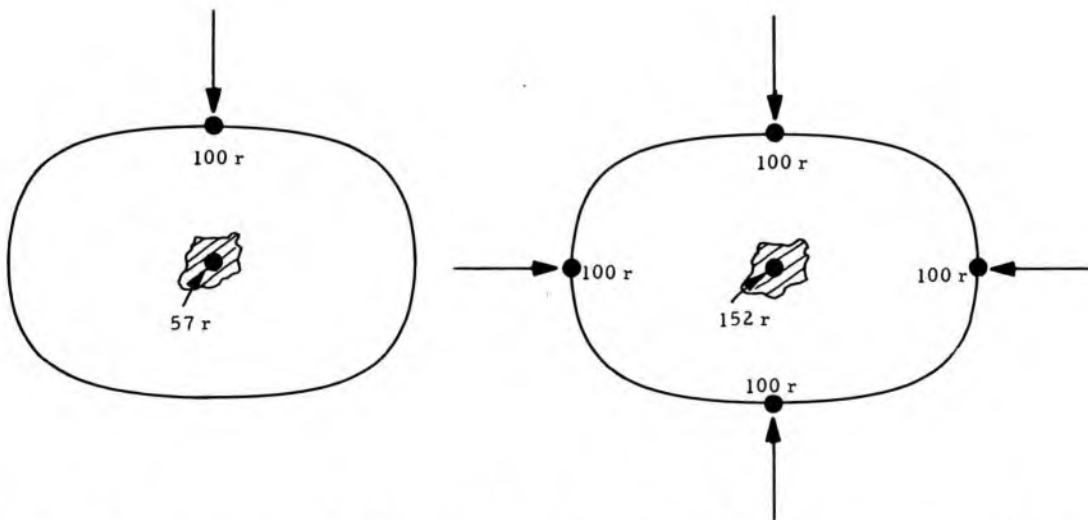


Fig. 8. Single vs. multiple portal treatment of a deep tumor. In this example, a central tumor in a 20 x 30 cm female pelvis is treated with cobalt-60 radiation at 80 cm source-skin distance with typical size portals. The arrows indicate central rays of each portal. Left. Single portal. The beam intensity is reduced in getting to the tumor, so only 57 rads are delivered centrally for 100 rads to superficial structures. The excessively high superficial dose effectively limits the tumor dose to sub-lethal levels. Right. Multiple portals. A more favorable tumor dose is delivered, because the beam is directed at each superficial area only for a fourth of the total treatment.

Theoretically, the most effective way to deliver maximum dose to a tumor with minimum dose to surrounding tissues is to insert the source of ionizing radiation *inside* the tumor. This cannot be accomplished using an x-ray machine, which must remain outside the patient. However, small radioactive sources *can* be placed in tumors and very frequently are. Some rapidly decaying radioactive materials may be left in the tumor permanently, but most sources (including those of radium) are left in place for the required time and then removed.

The above procedures all constitute efforts to increase the tumor dose relative to that received by surrounding tissues. A more subtle approach is to try to increase the tumor response to x-rays more than that of surrounding healthy tissues. This favorably alters the biologic situation we are trying to handle. Two types of quantities can be varied to achieve this differential response. The first involves chemicals. These may be both pharmacologic agents currently under development, and oxygen, which is administered either by regional perfusion or by the patient's breathing it under high pressure. The second method, fractionation, is older and simpler to use because it involves only the timing of successive treatments. These and other approaches will be discussed in Chapter 5.

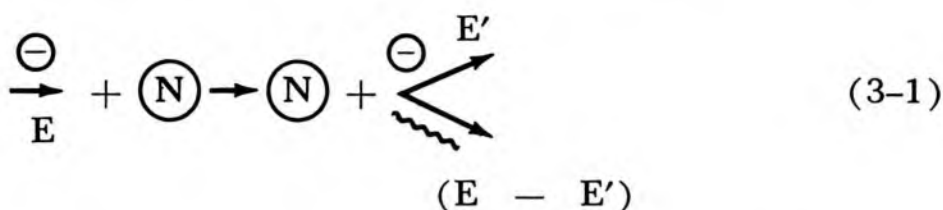
PRODUCTION OF X-RAYS

X-rays are photons with energies in the range of 1 keV to 35 MeV and above. High-energy radiation of this type can be produced as bremsstrahlen, characteristic, and gamma rays. It is important to clearly distinguish these three processes.³

1. **BREMSSTRAHLEN, OR "BRAKING-RAYS."** These can be produced when energetic charged particles such as electrons interact with atomic nuclei. Because electrons and protons are oppositely charged, they attract each other so the electron is deflected by the nucleus in much the same way as a comet is deflected by the sun. If the distance of closest approach is relatively great, the electron emerges with practically all its original energy and is merely deflected. This is called an "elastic" collision and is analogous to a very good golf ball's bouncing off a heavy concrete wall with virtually no loss in speed. However, suppose the electron comes much closer to the nucleus. The interaction then may be *inelastic* in that the electron gives up some of its energy and emerges with reduced speed (like a "dead" tennis ball). The lost energy appears as a photon. Such photons are called bremsstrahlen, a term de-

rived from the German: *bremsen*—to brake or slow down and *Strahlen*—rays. (Electrons are braked or slowed down by the interactions, producing *Bremsstrahlen*, analogous to the squeal of automobile brakes.)

A description of this process might be as follows:



Here $\textcircled{\text{N}}$ is used to indicate the nucleus (charges omitted), \ominus is the electron, and ~~~~~ represents the photon produced. Note the photon acquires the energy lost by the electron.* This mechanism is most important in the x-ray tube, so we shall have more to say about it. Bremsstrahlen† are also produced by charged particles in other situations, and can constitute a safety hazard. (See Chap. 14.)

2. **CHARACTERISTIC X-RAYS.** Unless electrons travel very fast, they tend to interact primarily with orbital electrons to produce ionization and excitation of atoms, as previously described. The atoms give up their energy later by restoration of the orbits which were temporarily deprived of their normal complement of electrons. When this occurs, the energy appears as one or more photons of discrete energy values. The energy of each photon is characteristic of the atom involved and the energy level interval through which the electron fell to reach the orbit location.

If restoration occurs in the outer orbit, photons of a few eV result (sometimes visible light); when inner orbits are restored, x-ray photons are produced. These x-rays can range from very low energies for biologic materials (0.28, 0.52, and 1.05 keV for carbon, oxygen, and sodium, respectively) through a few keV for light metals (up to 69.1 and 115 keV for heavy elements tungsten and uranium, respectively).

3. **GAMMA RAYS.** These photons accompany various nuclear reactions like radioactivity. Their energy generally ranges from a few keV through a few MeV. (See Chapter 9.)

* The nucleus actually acquires a small amount of energy. However, nuclei are so much heavier than electrons that this energy is negligible, analogous to the energy given an automobile when an insect bounces off the windshield.

† The term *Bremsstrahlung*, meaning braking radiation, is often used alternatively to *Bremsstrahlen*, braking rays.

Production of X-rays in a Tube

Both bremsstrahlen and characteristic x-rays are produced in x-ray tubes. Before discussing these rays in detail, we first shall briefly review what is needed to generate x-rays and describe the events involved in a general way.

FUNCTIONS OF THE X-RAY TUBE. Figure 9 shows the basic essentials for producing x-rays in a tube. These are as follows:

1. *An electron supply.* A heated tungsten filament (F) provides electrons. As previously described, the electron supply is controlled by means of an adjustable filament voltage (V_f); varying V_f changes the filament temperature and hence electron emission.

An electron focusing system is indicated symbolically in the drawing by the dotted line FS. This assures that a relatively small spot on the anode is struck by electrons.

2. *A high voltage supply; preferably dc.* This is impressed between the filament (F) and the anode (A) in order to ac-

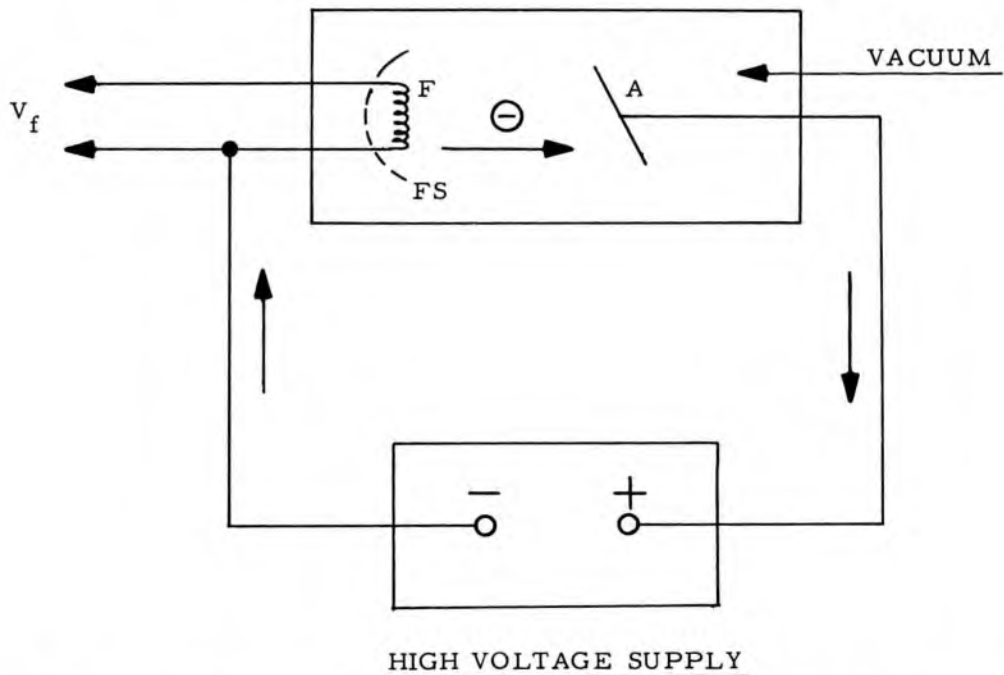


Fig. 9. Essential setup for controlled x-ray production. Items needed: F—Filament to supply electrons when heated by V_f . V_f —Filament voltage supply to heat filament to incandescence, thereby liberating electrons. FS—Focusing system to direct electron beam to local area of anode. A—Metal anode structure struck on small area by electron beam. High voltage supply (preferably dc) is provided to accelerate the electrons to desired high velocity. Vacuum in tube to permit unimpeded travel of electron beam.

celerate the electrons. It is usually made both adjustable and reproducible for proper control of the resulting x-ray beam.

3. *An anode structure (A)*. In most tubes the anode includes a part, usually made of tungsten, called the target (not shown). This is specially designed to withstand the collision of fast electrons.
4. *An evacuated container*. A high vacuum is required to minimize the production of gas ions by the electron beam. An excessive number of such ions destroys the ability of the tube to focus and otherwise control the electron beam. Loss of vacuum is a major cause of tube failure.

WHAT HAPPENS AT THE TARGET? Now let us consider in more detail what occurs at the instant electrons strike the target. The full story requires an accounting of all items involved when this collision takes place.

1. *Fate of the electrons*. The electrons are stopped almost instantly by the target. However, they survive this collision unscathed; they simply continue to be a part of the electrical circuit whose electron flow is indicated by the arrows in Figure 9. The electrons are essential to the production of x-rays but contrary to some misconceptions do not *become* x-rays.
2. *Fate of the electron's kinetic energy*. The electrons give up virtually all their kinetic energy during the collision. This is the energy they acquire as a result of falling through the high voltage between the filament and anode; it reaches a maximum just before collision.

At ordinary pkV values, this released energy serves primarily to heat the target. However, a small part appears as x-rays. The actual amount produced depends upon two things. The first is the chemical nature of the target: the x-ray intensity is proportional to the target atomic number. The second is the tube kilovoltage. The higher the kV, and hence the energy of the colliding electrons, the greater the x-ray production. Thus, the efficiency of x-ray production in the diagnostic region (150 pkV and below) is usually less than one percent (i.e., over 99 percent of the energy of the electrons ends up as heat). The efficiency is more than one percent in the orthovoltage region (180 to 300 pkV), and several percent at 2 MV constant potential. It rises rapidly above 2 MV, approaching 100 percent near 22 MV.

3. *Types of electron interactions*. A wide range of electron interactions is possible. The most common is a marginal interaction of the electron and an atom, which transfers only a small

amount of energy, simply causing the material to get warmer (nearly elastic collisions). However, the most useful interactions from our point of view are those yielding x-rays. In these interactions, the electron approaches the nucleus relatively closely and leaves the encounter deprived of a substantial part of its energy. This appears as a single photon, which has substantially all the energy the electron lost. Theoretically, an electron can interact with a nucleus in such a way that virtually all its energy appears as a single photon, but this is very rare.

4. *The photon versus electron energy.* It should be stressed that a photon is produced from only a single collision. Thus, two electrons *cannot* collide with the same nucleus to yield a single photon of higher energy than that of the more energetic electron. As a result, no matter how hard one may try, he cannot get high energy bremsstrahlung photons without having equally high energy electrons to produce them. In general, the energy of most photons is well below that of the electrons striking the target. Bremsstrahlen consist of photons whose energies range from zero to the original maximum electron energy.
5. *The fate of target atoms.* The production of bremsstrahlen is a result of electrons interacting with nuclei of target atoms. This slows the electrons down and yields photons. However, the atoms are essentially undisturbed in this interaction. The electrons can also interact with orbital electrons of the target atoms, rather than with nuclei. Under these circumstances, the anode atoms may be ionized and excited.

In summary, target atoms may receive glancing blows; these simply produce heat. They may be ionized and excited; their missing electrons are quickly restored with the production of characteristic photons. Finally, the electrons may interact directly with the nuclei, producing bremsstrahlen but leaving the atom itself intact.

6. *Heat production—significance.* As shown above, substantial amounts of heat may be produced in the target during the production of x-rays. This raises its temperature, particularly at the site of electron impact, called the electron focal spot. In diagnostic work the need for very short exposures usually requires that the x-ray tube target be operated at very high temperature.

Tungsten metal is used for almost all medical and industrial radiographic x-ray tube targets, partly because its high atomic number assures good efficiency of x-ray production. In addition, it has two

thermal advantages: first, a high melting point and second, a low vapor pressure, which prevents excessive volatilization of target material.

X-ray Bremsstrahlung Spectra

DEFINITION OF TERMS. The term “spectrum” refers to the relative amounts of different energy x-ray photons in the beam.⁴ This is of great practical importance because the photon energy distribution determines the penetration and other properties of the beam. A spectrum is analogous to a population distribution graph—i.e., numbers of people versus their age group. Figure 10(A) shows such a spectrum

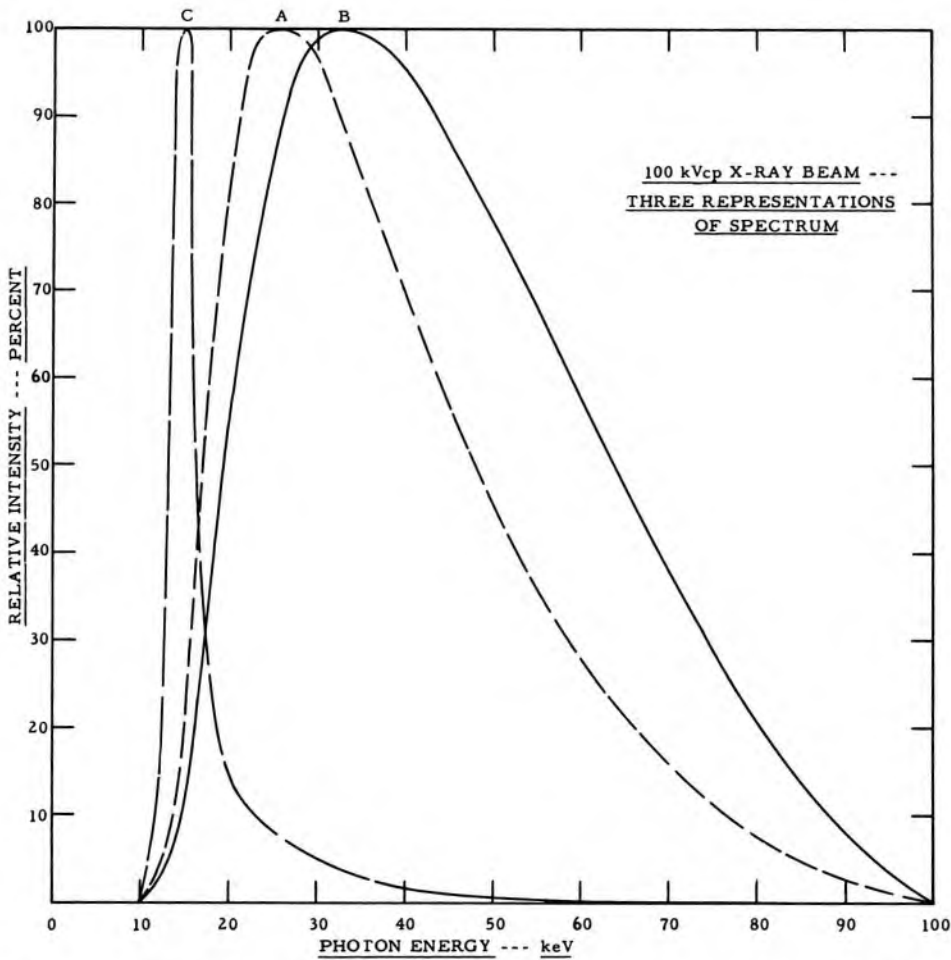


Fig. 10. Bremsstrahlung spectrum of a 100 kV (constant potential) x-ray beam. Three alternative representations: A. *Photons per second* impinging perpendicularly on one cm² area, plotted vs. photon energy. This is a population profile of the photons. B. *Total photon energy per second* impinging perpendicularly on a one cm² area, plotted vs. photon energy. This is the classic definition of intensity (solid curve) and is used throughout this book. C. *Roentgens per minute* at a location, vs. photon energy. This is what interests the radiologist in radiotherapy and radiation protection and is properly called exposure rate.

of an x-ray beam. Note that the abscissa is photon energy, the ordinate "relative intensity" (maximum arbitrarily called 100 percent).

The term "intensity" can be somewhat ambiguous because it means different things to persons in different disciplines. Strictly speaking, the term should be applied only in the basic physical sense: the energy incident perpendicularly each second on a 1 cm² area. However, a radiation physicist using a modern solid state detector spectrometer measures the number of photons perpendicularly incident each second on 1 cm² of a detector [Fig. 10(B)].* These are not the same. For example, consider 1,000 photons arriving each second on the detector. If these are each 1,000 keV, energy is received ten times faster than if they are 100 keV, even though the arrival rate is the same in both cases. The *energy* curve (B) is hence shifted to the right of the *photon number* curve (A) in Figure 10.

To complicate the picture further, radiation therapists normally think of intensity as the number of *roentgens* (a unit of x-ray exposure, Chap. 6) *per second* reaching the patient. This is the exposure rate, which differs in nature from both of the other types [Fig. 10(C)]. While it is too early to define the roentgen specifically, it must be stressed that the spectrum will be shaped differently when drawn using one of these definitions than if drawn using another.

In all cases, we use the term "relative" intensity simply to be able to ignore the fact that getting closer to the source or increasing tube mA increases the intensity. When we speak of spectrum here we are interested only in the general shape of the curve, so we arbitrarily call the maximum height of the spectral curve 100 percent.

In all the spectra to be shown below, we shall use the energy arriving perpendicularly per cm² per second to define the ordinate. Actually, by standard procedures, one can easily convert from this representation to the other two if desired.

The spectrum is the most accurate representation of x-ray beam quality or photon composition. However, in practice it is both impractical and unnecessary to measure spectra of x-rays from each of the tens of thousands of x-ray units in use, for each kV and filter used. For most medical work, a simpler quantity is used which is far easier to measure. This is the half-value layer (HVL), sometimes expressed as "half-value thickness" (HVT). In this book, the term HVL is retained. It is defined as the thickness of an appropriate absorber which reduces the intensity of a narrow portion of the beam in question by 50 percent. It is relatively easy to measure and gives a useful idea of beam quality, so it is almost universally employed.

* Properly called the "particle fluence rate" or "particle flux density."

BREMSSTRAHLEN SPECTRUM VERSUS OPERATING FACTORS. This spectrum depends on both absorption and kV considerations.

Figure 11 shows computed spectra produced at three locations when a constant dc voltage is impressed across an x-ray tube. Here Kramer's simple formula is used, and, while it is only approximate, it provides a fairly good result and is useful for this general presentation.* The first location is right in the target, where the x-rays are initially produced (A). This spectrum is a straight line sloping downward toward the keV axis. There is a maximum photon energy equal to the energy of the electrons striking the target. The high energy photons contribute relatively little to the total x-ray intensity, while other photons contribute increasingly more as we go down in photon energy.

Before photons can leave the x-ray tube they must traverse material in the way, such as the tube glass envelope, housing oil, and window. These intermediate materials absorb some photons of all energies. However, the less penetrating lower energy photons are removed disproportionately by this intervening material, and the second curve results (B). An additional absorber (a filter) is usually inserted in the beam; consequently the beam striking the patient contains even fewer lower energy photons (C).

Three particular photon energies are of interest in describing the spectrum of an x-ray beam. (Fig. 12). The first is the maximum photon energy E_{\max} . As pointed out above, this is numerically the same as the maximum electron energy of the electron beam striking the target. In general, it is determined by the peak kilovoltage applied across the tube, for this defines the maximum electron energy. Thus, a 100 pkV tube voltage results in 100 keV maximum photon energy. At the other end of the spectrum lies E_{\min} , the minimum photon energy. While this is less uniquely determined than E_{\max} , it generally is much the same for any operating kilovoltage, depending primarily on the total filtration to which the photons are subjected. Between this maximum and minimum pair of values, lies E_{eff} , or the effective energy. This is defined as the energy of a hypothetical beam with only one kind of photon (called a "monochromatic" beam), having the same half-value layer as the actual beam under consideration. There is no simple general relationship between the E_{eff} and E_{\max} values.

It will be recalled that photon wavelength and energy have the relationship $\lambda = 12.4/E$. Hence, for each of the above photon energies there exists a unique wavelength. These have been indicated in parentheses beneath corresponding photon energies in Figure 12.

* Kramer's Law: $I = K (E_0 - E)$, where I = intensity of photon with energy E , k is a constant, and a constant potential E_0 is assumed.

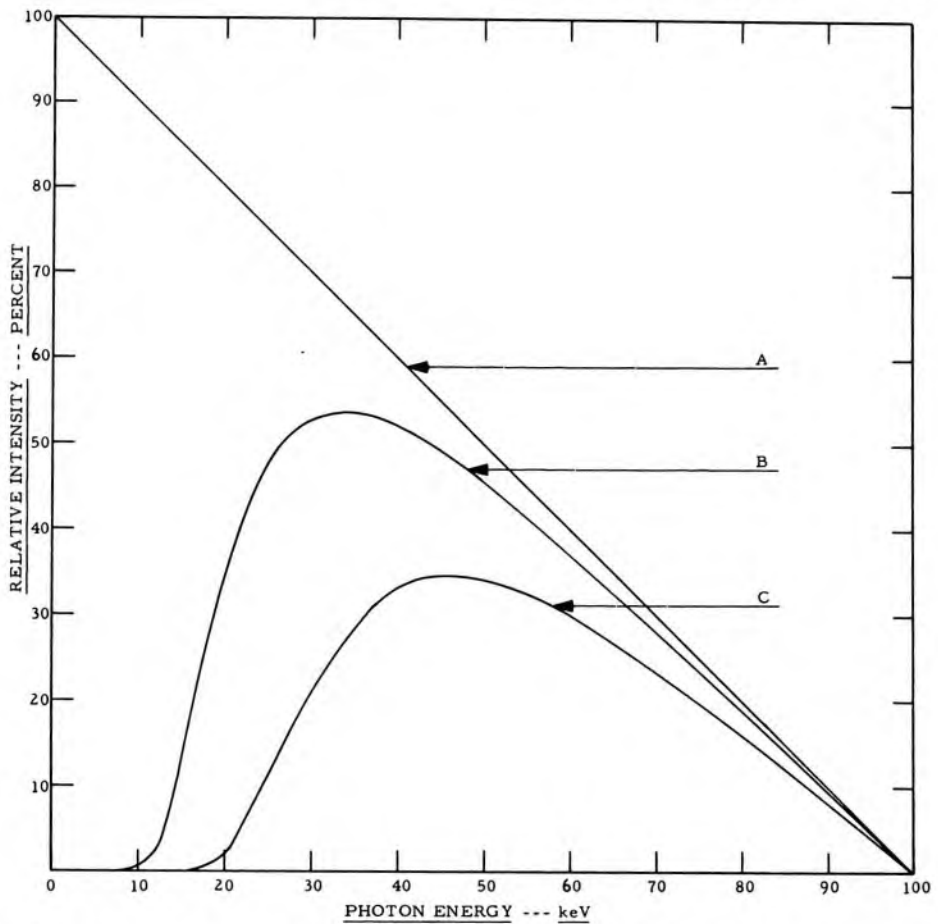


Fig. 11. X-ray spectra at three locations—100 kVcp (constant potential). A. As produced within the target, before attenuation on the way out of the tube. B. In beam leaving tube, after traversing the inherent filtration of the tube and its housing. C. In beam reaching patient, after traversing both inherent filtration and external filter.

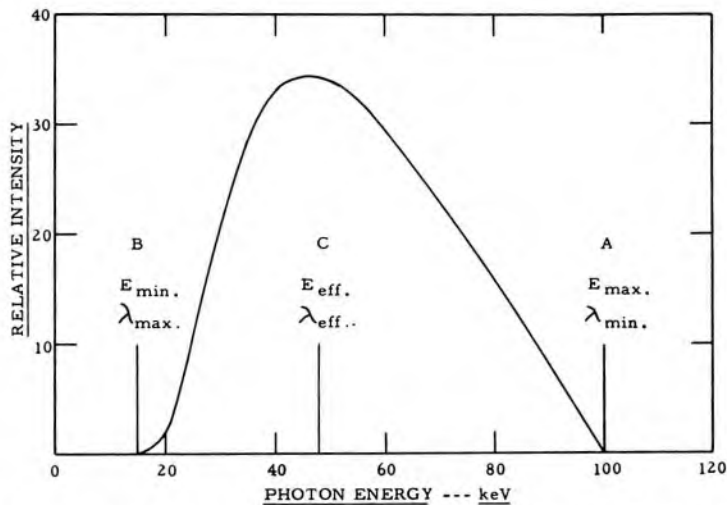


Fig. 12. Definition of particular photon energy values. A. *Maximum energy, minimum wavelength*, determined by operating pkV. B. *Minimum energy, maximum wavelength*, determined by total filtration. C. *Effective energy and wavelength* depends in a complex way on pkV, filtration, and tube voltage wave form. Closer to minimum vs. maximum energy.

Figure 13 shows the effect on a spectrum of varying the tube kilovoltage. Note that increasing kilovoltage from 80 to 100 kV simply shifts the entire spectral curve to the right, adding primarily high keV photons.

It is evident that a small increase in kilovoltage greatly increases the number of high energy photons, thereby considerably "hardening" the beam. Correspondingly, lowering the kilovoltage greatly reduces the high energy photon population in the beam, greatly reducing its penetration. This fact is employed routinely in all clinical x-ray work.

Figure 14 shows the effect of varying filtration on the transmitted x-ray spectrum. Note that when the filtration is increased, lower energy photons are more effectively removed than high energy photons, and the left edge of the curve is consequently shifted toward the right. Unfortunately, some of the higher energy photons are also removed so that this type of purification of the beam cannot be continued indefinitely with profit. A point is ultimately reached where any fur-

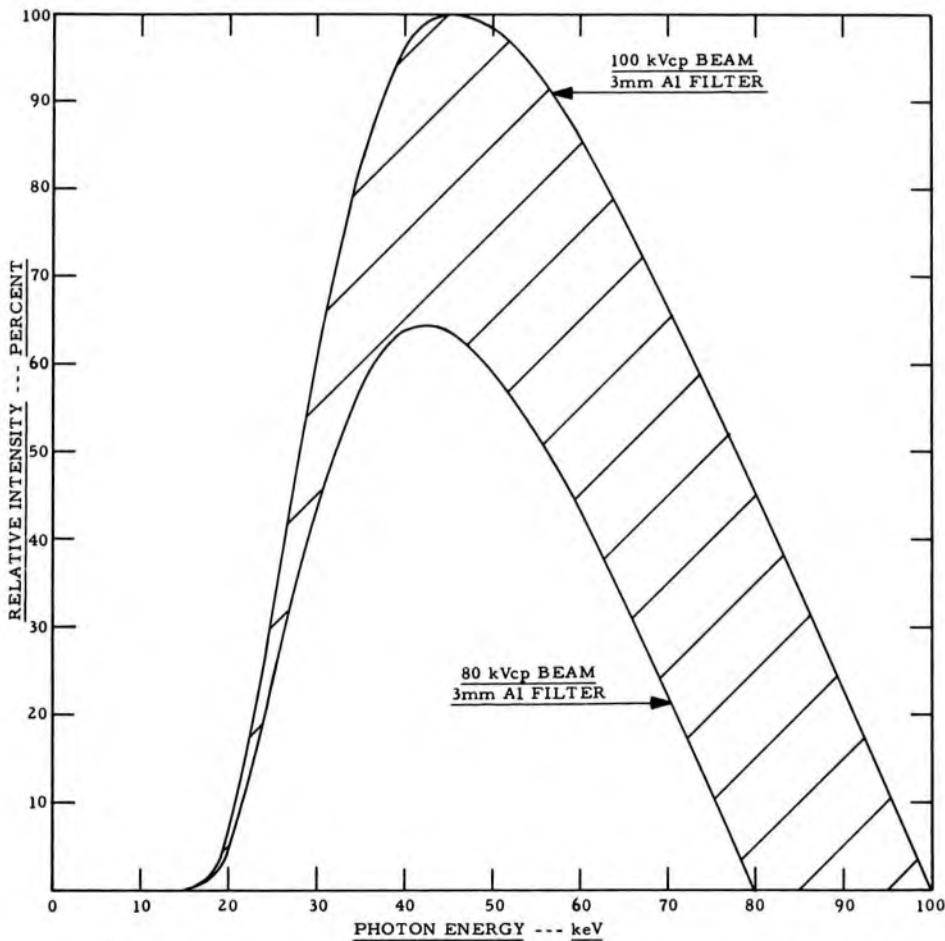


Fig. 13. The effect on spectrum of increasing kV—3 mm Al added filter. Note that increasing kV from 80 to 100 kVcp hardens the beam greatly, by adding *high* energy photons disproportionately to low energy ones. Note that the x-ray beam intensity is considerably *increased*.

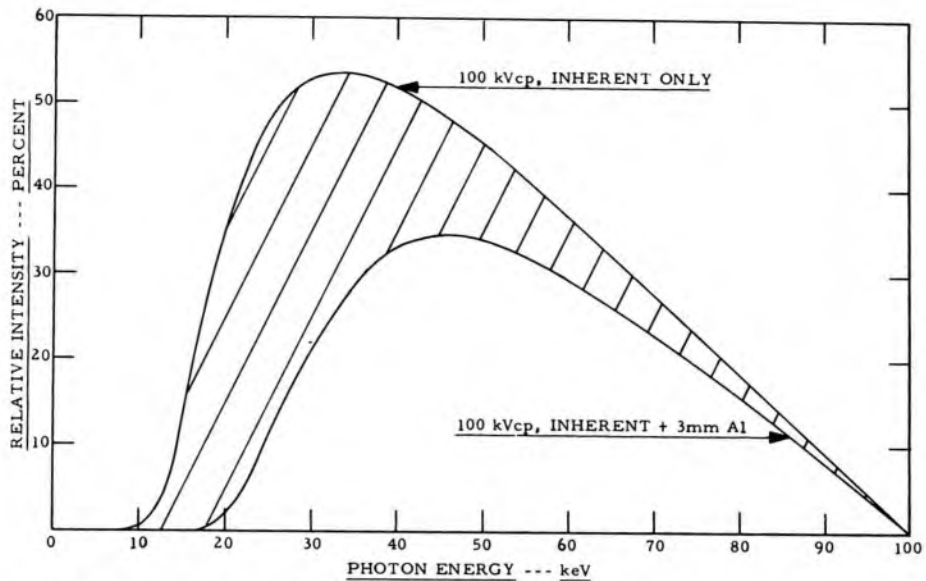


Fig. 14. The effect on spectrum of increasing filtration—100 kVcp. Note that adding to the inherent filtration (1 mm Al equiv. assumed) *hardens* the beam. The mechanism, however, is different from that of Figure 13 since it involves the selective *removal* of *low* energy photons. Hardness is increased at the cost of *reducing* total beam intensity.

ther filtration reduces the intensity of the desired high energy components to such an extent that further filtration is uneconomical. In general, the choice of filtration represents a compromise between adequately great E_{\min} and E_{eff} values and reasonable beam intensity.

The spectra discussed so far have been those produced by a dc constant potential voltage. In practice, pulsating dc voltages are employed in most diagnostic x-ray applications. Figure 15 shows what happens to the shape of the spectral curve when one changes the tube voltage from dc constant potential to a *pulsating* dc of the same peak value (rectified wave form). Two effects are noted. First, the total intensity is considerably reduced, roughly by half in the example shown. Second, the greatest loss is in high energy components: for the same maximum voltage, the effective photon energy using pulsating voltage is considerably less than that using constant potential. This is true in both diagnostic and therapeutic x-ray beams, and even at voltages in the millions of volts. For this reason it would be theoretically advantageous to use constant potential voltage generators in radiology; however, such generators are more costly and complex, so pulsating x-ray machines are used more generally in diagnostic work, as well as in several therapy units.

WAVELENGTH TYPE OF SPECTRUM GRAPH. Historically, x-rays were first studied by physicists investigating spectra obtained by crystal

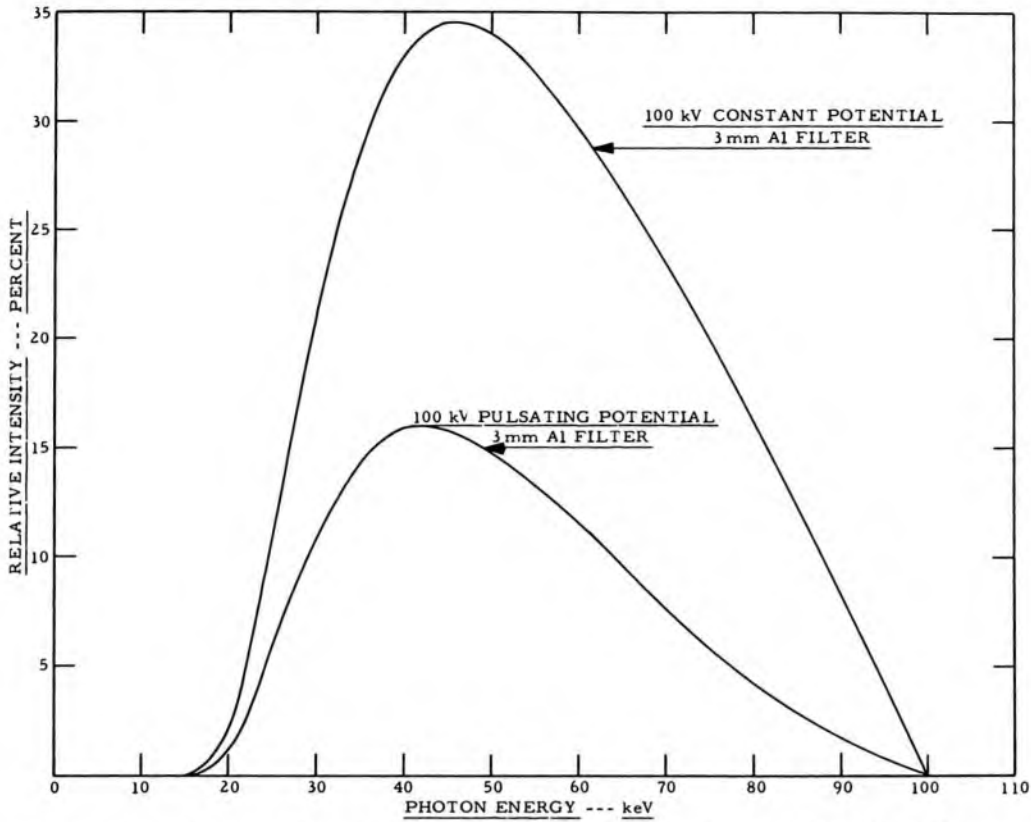


Fig. 15. Comparison of spectra produced by pulsating high voltage (100 pkV) vs. constant high voltage (100 kVcp)—3 mm Al filter. Note the constant potential beam has roughly twice as much intensity as the pulsating potential beam. In addition, it is also somewhat harder, having proportionately more high energy photons present.

diffraction. In such experiments the most easily measured quantity is wavelength. As a result, the original spectra were plotted with photon wavelength rather than energy as the horizontal coordinate, and even recent literature often shows spectra plotted this way.

Figure 16 compares spectrum graphs of the same x-ray beam plotted two ways: with keV and wavelength abscissas. Note that the curves look entirely different. In our discussion, we have stressed the photon energy type of drawing for two reasons. First, newer data are obtained using scintillation and similar spectrometry methods; they are almost universally plotted as intensity versus photon energy. Second, the wavelength diagram gives an erroneous impression as to the amount of x-ray energy in the low as compared with the high energy part of the spectrum. In Figure 16(A), it is evident at a glance that an equal amount of energy lies below about 52 keV as above. This gives the reader a feeling for the general quality of x-rays emerging from the tube and striking the patient. Fifty-two keV corresponds to 0.238 Ångstrom, approximately, and it is clear from even casual in-

spection of Figure 16(B) that the areas to the left and right of this line are not identical. The wavelength representation therefore distorts the picture and can lead to incorrect inferences as to relative amounts of high and low energy photons in the beam, unless corrected for scale distortion.

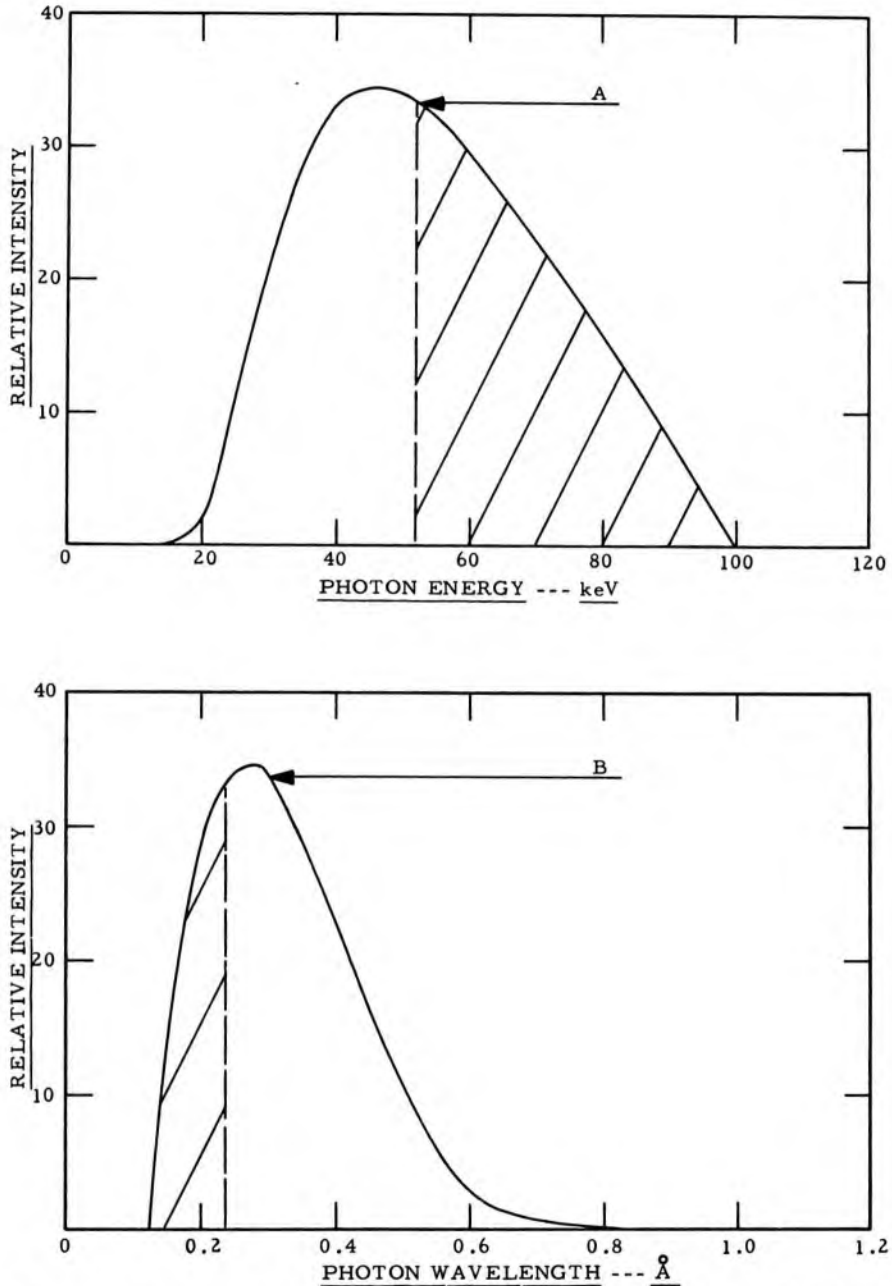


Fig. 16. Comparison of photon energy vs. wavelength method of plotting the same x-ray spectrum. Both graphs represent the same x-ray beam, generated with 100 kVcp, 3 mm Al filter. In both representations the shaded area represents the half of the total spectrum energy with higher energy photons. Curve A clearly shows half the beam energy is delivered by photons 52 keV and greater. Curve B gives a distorted impression of mostly *low* energy (long wavelength) photons in the beam.

Characteristic Spectra

Characteristic x-rays are also produced at the x-ray tube target because the electron beam ionizes target atoms. Unlike bremsstrahlung, the characteristic photons have a unique, rather than a continuous range of energy values. For example, those from tungsten targets have values of 69.1 and 11.3 keV, with a few additional values slightly lower as well. Characteristic spectra are sometimes also called line, homogeneous, discontinuous, and monochromatic, in contrast to broad, inhomogeneous, continuous, and polychromatic corresponding terms for bremsstrahlung.

In clinical practice, tungsten characteristic K-orbit photons represent a relatively small part of the total x-ray beam reaching the patient, and exert no special practical effect. However, the 11.3 keV photon (as well as a few below this, down to about 8 keV) may be of considerable importance in very low energy x-ray work. (For example, applications include superficial therapy with low filtration as well as mammography and similar applications.)

An interesting point should perhaps be stressed regarding the production of characteristic radiation. The 69.1 keV photon of tungsten is produced after an atom loses a K-orbit electron from collision with an incoming electron. It is hence evident that ionization of the K-orbit is a prerequisite for the production of this K-orbit characteristic radiation. For this reason, if the operating kilovoltage is not sufficient to provide a 69.5 keV electron beam, no characteristic photon of this energy can be produced.* Furthermore, the amount of characteristic radiation produced rises rapidly as the operating kilovoltage is raised above the threshold value.

In x-ray diffraction work, characteristic radiation is most useful for carrying out the analysis. Therefore, the target is made of the material whose characteristic radiation is desired, and very low filtration is employed to permit this soft radiation to reach the material being irradiated.

X-ray Tubes

The x-ray tube is the heart of the machine. Improper operation may result in destruction of the tube itself, costing from a few hundred dollars to several thousand dollars, depending on its type. In addition, tube failure is often accompanied by damage to other expensive high voltage components, and lost operating time is both an-

* The binding energy of the K-orbit electrons is 69.5 keV. Only 69.1 keV photons are produced because in heavy atoms the K orbit is preferentially restored by electrons from outer orbits, rather than by free electrons.

TABLE 3(A). ROTATING VS. FIXED ANODE X-RAY TUBES

General Data and mA Ratings

Tube Type ¹	Focal ² spot— mm	Max. pkV	Anode		Allowed mA at 90 pkV full wave, for exposure times in sec of:		
			H.U. ³ cap.	Cooling ⁴ Rate	0.1	0.5	2.0
Dynamax-40	1.0	125	110,000	30,000	225	180	135
Dynamax-150	1.0	150	135,000	45,000	280	250	205
Dynamax-60-60 cycles	1.0	150	200,000	54,000	340	280	220
Dynamax-60-180 cycles	1.0	150	200,000	54,000	560	450	320
Dynamax-40	2.0	125	110,000	30,000	430	350	250
Dynamax-150	2.0	150	135,000	45,000	470	450	300
Dynamax-60-60 cycles	2.0	150	200,000	54,000	730	570	400
Dynamax-60-180 cycles	2.0	150	200,000	54,000	1,000	830	510
CYSL	2.3	100	92,500	35,000	83	68	55
SLT	2.3	140	200,000	102,000	85	67	56
CYSL	4.3	100	92,500	35,000	200	150	120
SLT	4.3	140	200,000	102,000	200	150	120

Notes:

¹Tube types: "Dynamax" are Machlett rotating anode tubes. Others are Machlett fixed anode diagnostic tubes.

²Focal spot is square area of indicated length each side.

³Heat units (H.U.) = #pkV × #mA × #sec. during exposure.

Capacity is total accumulated heat units limit, beyond which anode assembly damage may occur.

⁴Cooling rate in heat units per minute. Figure is maximum rate occurring when anode is hot as permissible. Normally, cooling rate is considerably less.

TABLE 3(B). CONVENTIONAL THERAPY X-RAY TUBES

Tube Type	Maximum pkV	Cooling Method	Typical Operation
Aeromax 12, 15*	110 pkV	Air circulator	100 pkV, 4 mA
Thermax	140 pkV	Water-cooled	140 pkV, 8 mA <i>or</i>
Thermax	140 pkV	Water-cooled	110 pkV, 10 mA
Thermax	140 pkV	Oil-cooled	140 pkV, 12 mA <i>or</i>
Thermax	140 pkV	Oil-cooled	110 pkV, 15 mA
DT-220	220 pkV	Oil-cooled <i>only</i>	220 pkV, 20 mA <i>or</i>
DT-220	220 pkV	Oil-cooled <i>only</i>	200 pkV, 25 mA <i>or</i>
DT-220	220 pkV	Oil-cooled <i>only</i>	200 pkV constant potential, 18 mA
DT-260 k	260 pkV	Oil-cooled <i>only</i>	260 pkV - 18 mA

*Note: Machlett tube numbers listed. Similar units available from other manufacturers.

Tables courtesy of Machlett Laboratories, Inc.⁸

noying and costly. Table 3 summarizes useful data regarding diagnostic and other applications of some representative x-ray tubes. A wide variety of tube types is available, depending on the required current, exposure duration, arise in special applications.⁵⁻⁸

BASIC DESIGN PROBLEMS. Certain basic problems are common to all x-ray tubes. They include provisions for voltage insulation, electron beam control, anode cooling, and other special problems. These are discussed below (see Fig. 17).

1. *Insulation.* High voltage breakdown can occur in several ways. The first is by arcing around the tube. This can be through the oil from either of the tube electrodes to the housing, which is of course at ground potential. The glass of the x-ray tube must insulate against the full voltage between anode

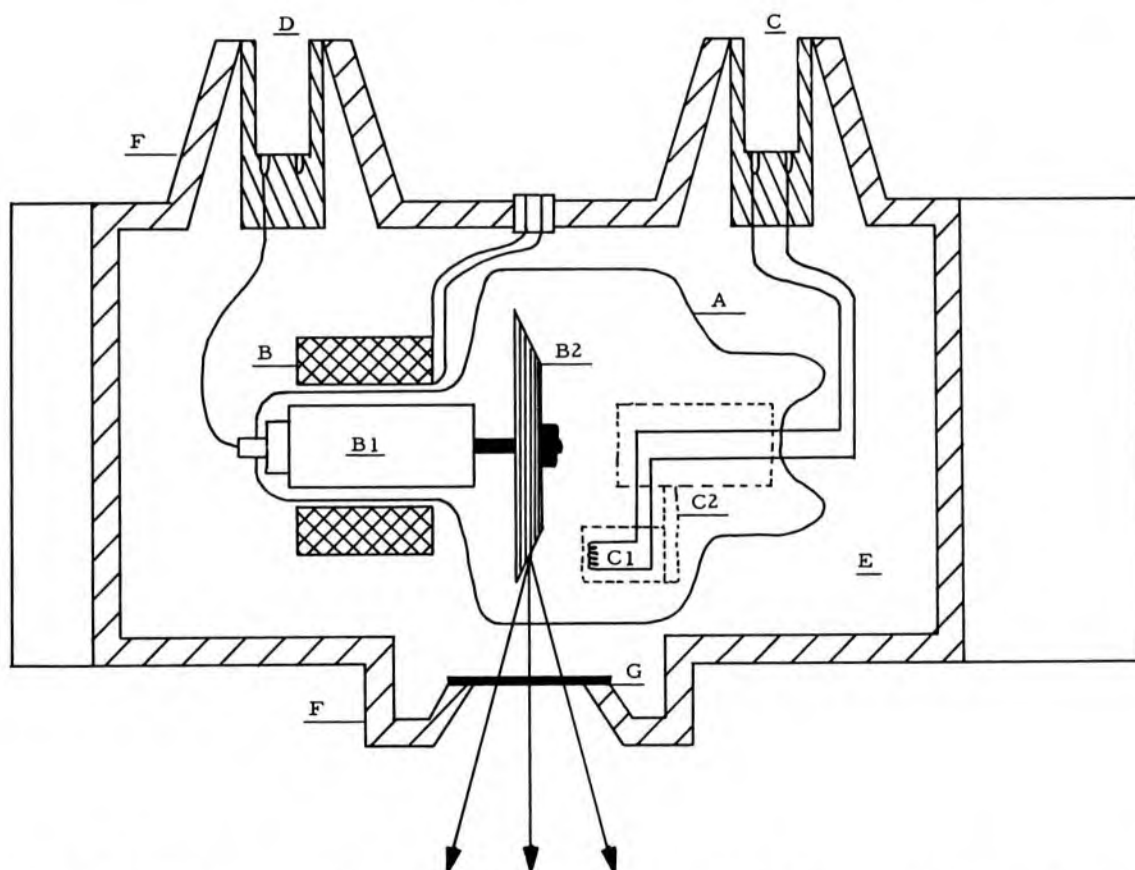


Fig. 17. Basic essentials of a diagnostic x-ray tube assembly (schematic only). A. X-ray tube; filament assembly at right, anode at left. (Anode armature mounted on special vacuum bearings not shown.) B. Field winding coil rotates armature (B1) and anode (B2). C. Filament cable socket: C1-filament, C2-filament shield and support (shown by dotted lines). D. Anode cable socket. E. Oil for insulation against high voltage. (Bellows provided at right end of housing for temperature expansion; not shown in drawing.) F. Metal housing with lead shielding enclosed (not shown in detail). G. Plastic oil seal for housing provides x-ray beam window. (Redrawn from Kirka.⁶ Courtesy of Machlett Laboratories, Inc.⁸)

and cathode; it must be properly annealed and of controlled composition to withstand the high voltage stresses. As the tube operates, metallic deposits gradually accumulate on the inner surface due to operation of metal parts at high temperature. (Careless or improper operation of tubes can result in premature occurrence of this phenomenon with resultant shortening of tube life.) Finally, the tube vacuum must be acceptable, of the order 10^{-5} mm Hg pressure, or less. While gases and vapors within the tube are in themselves normally nonconductive, any flow of current through the tube will ionize their molecules by collision. These ions produce currents where they should not flow (such as positive ions bombarding the hot filament). Furthermore, they upset the entire flow of electrons from filament to anode. The defocused and misdirected electron beams trigger events which can upset operation of and even destroy the x-ray tube.

As the tube is operated, its vacuum inevitably deteriorates with time, despite the presence of an active "getter."* This is because target heating inevitably results in removal of surface tungsten, uncovering volatile materials and gases previously confined. When released, these materials contaminate the vacuum. However, it must be emphasized that this release usually does not pose serious problems until well beyond guaranteed life if the tube is properly operated.

2. *Electron beam control.* The electron beam must strike the target rather than the glass envelope; in addition, it should impact only a small area of the target, especially in diagnostic tubes, to assure proper definition of the resulting x-ray beam. This is accomplished by both electrostatic focusing and special design of the filament and anode.

Figure 18 shows schematically the filament construction and electron beam flow in a "line-focus" type of tube used in all x-ray diagnostic work. Note that the filament is a single helix of rather heavy tungsten wire, about 1 cm long and 1.5 mm in over-all diameter. The electron beam is directed toward the anode where it strikes the target surface, which is inclined at a rather steep angle (usually about 15°). The area struck on the target (the electron or target focal spot) is even larger than the area of the filament itself. The x-rays travel in all directions, but the useful ones must emerge in the direction indicated. As "seen" by the object radiographed

* "Getter" is a deposit of alkali metal left inside vacuum tubes after sealing. It helps maintain the vacuum by combining with gas and vapor molecules, particularly hydrogen, oxygen, nitrogen, and water.

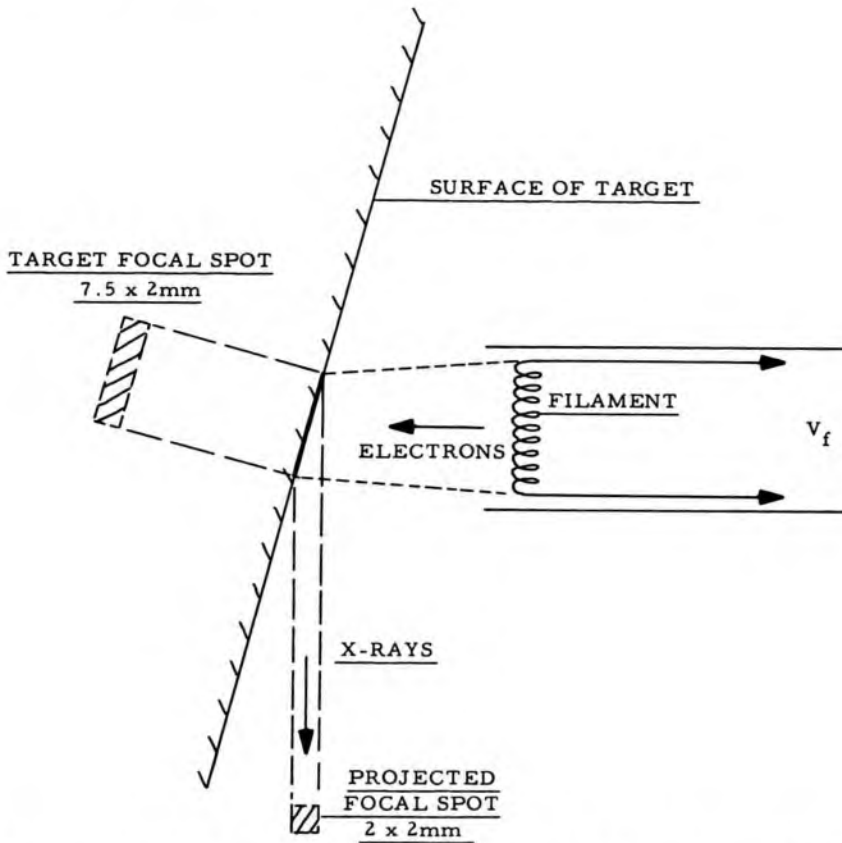


Fig. 18. The line focus principle in diagnostic x-ray tubes. The filament is a helix of tungsten wire much longer than wide. The electrons therefore strike a rectangular, rather than square or circular, area of the target. Because of the sharp slope of the target, the x-rays appear to originate from a square source, smaller in area than the actual electron or target focal spot. The net result is a smaller projected than target focal spot, making possible sharp roentgenograms with higher tube mA ratings.

and the film from below the tube, x-rays appear to emerge from a spot which is almost a square. This is referred to as the "projected focal spot." The effect of this design is to yield a small square *x-ray* focal spot while allowing the electrons to strike a larger area rectangular *electron* focal spot. The former helps to achieve sharp roentgen images; the latter, to dissipate the electron beam heat over a maximum target area, thereby permitting a greater mA rating for the tube.

In higher voltage x-ray therapy, special problems can arise from some electrons after they strike the target. At energies of 150 keV and over, many electrons are deflected by the atoms in the target and re-emerge with fairly high velocity. While most tend to return to the target, some may escape and strike various parts of the tube with undesirable results. Deep therapy x-ray tubes hence may employ a special

hood structure around the target to prevent these electrons from reaching the glass and other critical portions of the tube (see Fig. 19). A metal (beryllium) window of low x-ray absorption is employed in this hood; it permits the desired x-rays to leave the tube but readily captures scattered electrons.

3. *Anode temperature.* The target itself and the anode as a whole may be damaged by heat. This type of problem is probably the most difficult one to control in practice because overheating can occur by accident or negligence. However, many useful special designs have been worked out to help solve this problem, and they are of considerable interest to the user because of the important role they play (see below).
4. *Special problems.* The above cover the vast majority of problems. However, others arise in specific applications and merit brief mention.

It is sometimes desired to treat some very superficial lesions with the x-ray source placed very close to the patient's

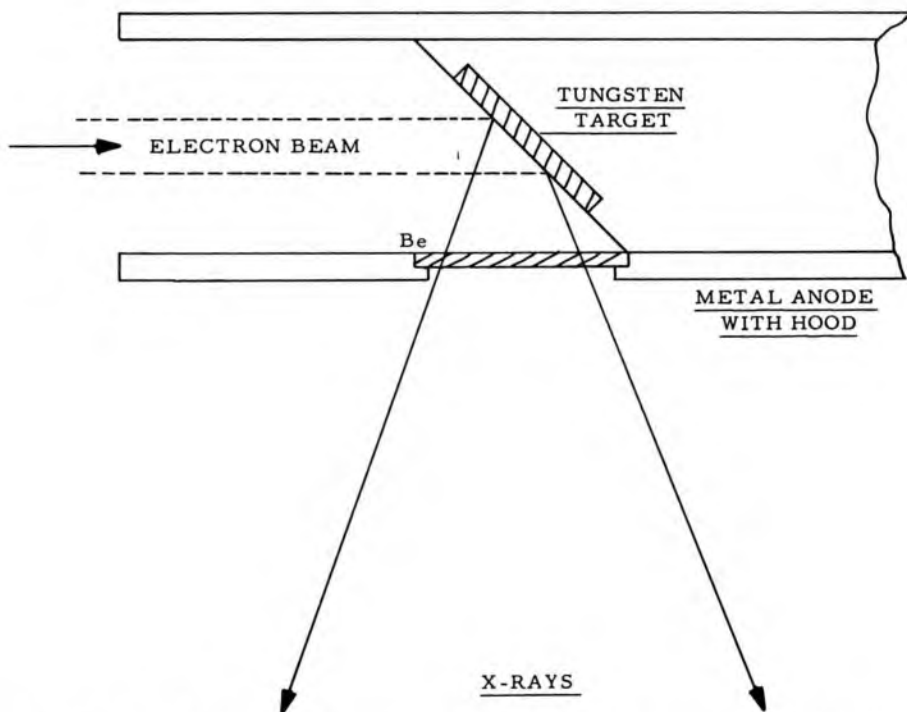


Fig. 19. Metal hood around target area to trap scattered electrons. Shown is a system used in some deep therapy tubes (150 to 250 kV) to trap electrons scattered out of the target. A metal hood structure at anode potential captures most such electrons before they can damage the glass envelope. A small part of the metal hood must be transparent to x-rays, so a beryllium insert is employed. Since beryllium is a good electrical conductor, its substitution does not reduce the hood's effectiveness.

skin. In one machine, this distance can be as short as 1.8 cm. In general, the distance for such so-called "contact therapy" ranges from 1.8 cm up to 5 cm. Such machines operate at kilovoltages 50 kV and below. They require that the tube window permit the passage of very low energy photons. To achieve this, tubes are made with windows of either very thin special glass or beryllium. It must be stressed that such "contact therapy" machines, as well as beryllium window tubes in general, produce extremely great x-ray intensities (in the range of several thousands of roentgens per minute and higher). *They therefore constitute a serious potential hazard to the operator and the patient.*

Another special application is x-ray diffraction. This work employs x-ray tubes with changeable targets, such as copper, nickel, etc. Characteristic radiation emitted by these targets is desired and is always of very low photon energy (of the order of 5 keV or 2.5 Ångstrom units). In addition to the provision for versatility in changing targets, these units must also operate with extremely transparent x-ray tube windows to permit very low energy photons to reach material being spectrographed. As a result, special procedures must be followed to prevent injury to the operator during alignment and other procedures.

FOCAL SPOT DAMAGE. The remainder of our discussion of x-ray tubes concerns specific anode heating problems and their solutions. Excessive heating can damage the target locally. This is a consequence of heat being produced at the focal spot faster than it can be removed by heat conduction and radiation at tolerable temperatures.

If the overheated anode is of the fixed type, molten tungsten rolls downward due to gravity. This leaves a depression at the electron focal spot with a solidified ridge of tungsten below [Fig. 20(Top)].

Most modern diagnostic tubes have rotating anodes, for reasons discussed below. In these tubes, the physical area of the target struck by electrons becomes an almost flat ring rather than a small rectangle, so no single depression is produced unless the anode fails to rotate. However, if this ring area becomes overheated, a generalized roughening or "pitting" occurs [Fig. 20(Bottom)]. Although not so serious as melting, pitting does reduce the available x-ray supply because the rough tungsten surface absorbs many photons on their way out of the target.

Localized target overheating produces two other adverse effects. First, tungsten is volatilized from the target area and may deposit

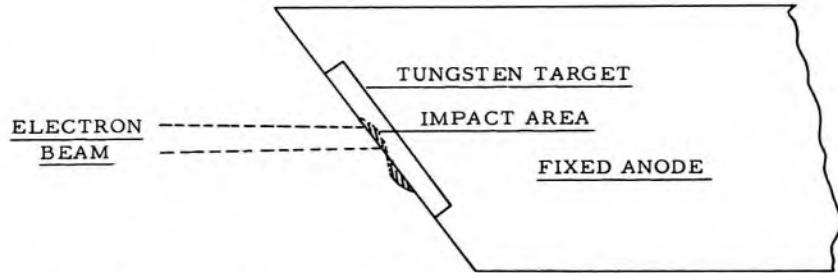
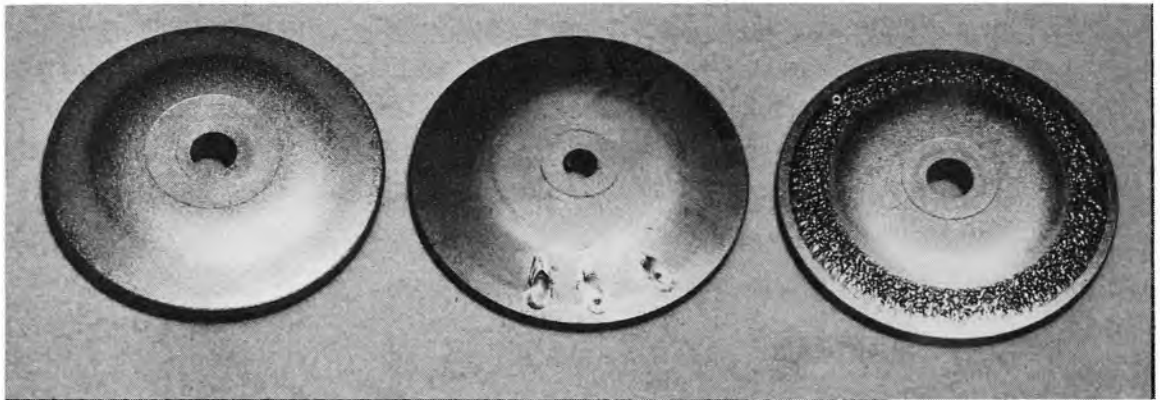


Fig. 20. Target damage from overheating. Top. Localized melting in fixed anode tube (effect exaggerated for clarity). The impact area is hottest, melting tungsten. This falls and solidifies in cooler area below. Bottom. Three targets removed from rotating anode tubes after use, from left to right: left—proper use, relatively undamaged; center—target melted by operation with stationary target; right—target pitting from excessive loading, with motion. (Bottom illustration courtesy of Machlett Laboratories, Inc.⁸)



on the glass surfaces of the tube. Second, removal of any metal from the target surface exposes virgin metal with accompanying release of gases and vapors; these tend to destroy the vacuum.

There is much less danger of target pitting with a large electron focal spot than with a small one. The reasons can be illustrated with the analogy of a man holding his hand above a burning candle. If he lets the flame steadily contact a single area of his palm, he burns himself. If he moves his hand above the flame heated areas are continually replaced by cool ones; the heat is dissipated over a larger area, and no one part of the hand receives enough heat to become intolerably hot. For this reason, use of a larger focal spot in an x-ray tube permits operation at higher mA values. [See Table 3(A).]

Modern radiography has created demands for tubes capable of safely delivering currents up to 1,000 mA, at 125 pkV and higher. Many studies require very short exposures to control *motion blur*. To illustrate, exposures as low as 1/120 second are required to assure a satisfactory radiograph of the chest of an infant or a very sick or apprehensive adult. During longer exposures, excessive motion of the

margins of important organs causes unacceptably blurred images. The effect of various degrees of motion on image sharpness and contrast is shown in Figure 21. Thus, high mA tube capacity is needed to prevent motion blur in many applications.⁹

Older tubes solved this problem simply by the use of relatively large focal spots permitting higher mA operation. Unfortunately, this simply replaces one problem with another because a large focal spot helps overcome motion blur at the sacrifice of increased *geometric or penumbra blur* (Chap. 8). Figure 22 illustrates the effect of penumbra blur.

Another expedient is to have small focus tubes for light duty and large focus for heavy duty work. This, however, leads to complexity, confusion, and delays unacceptable in a busy department. In addition, it still does not satisfy the requirement of high mA and

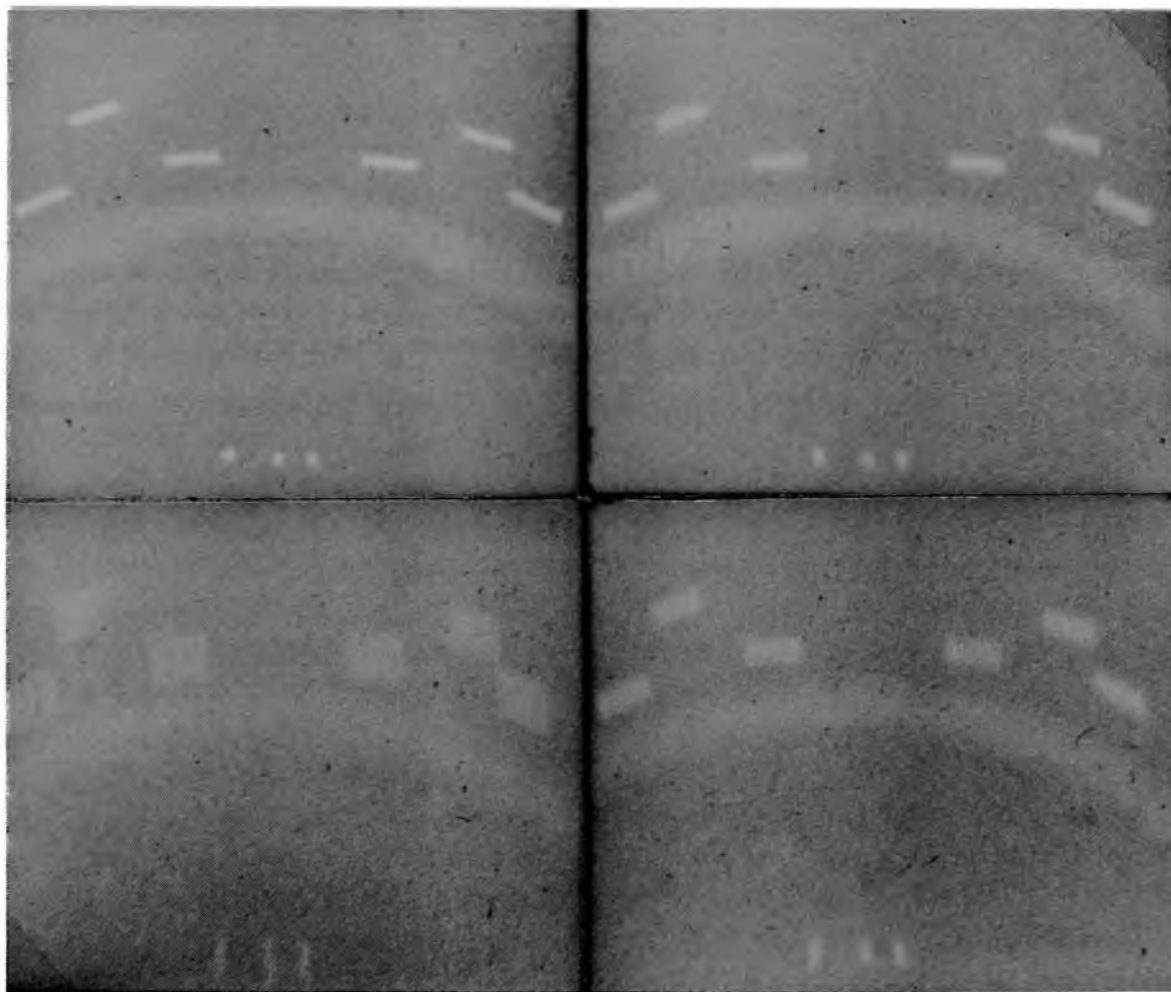


Fig. 21. Effect of motion on small x-ray images. Going clockwise, on photograph, total motion was 0, 0.5, 1.0, and 2.0 mm during the exposure. Small and narrow images—0.525 mm aluminum object. Arc band image—2.18 mm water object. Note contrast reduction as well as blur is produced by motion. (Redrawn from Stanton and Lightfoot. *Radiology*, 83:42, 1964.)

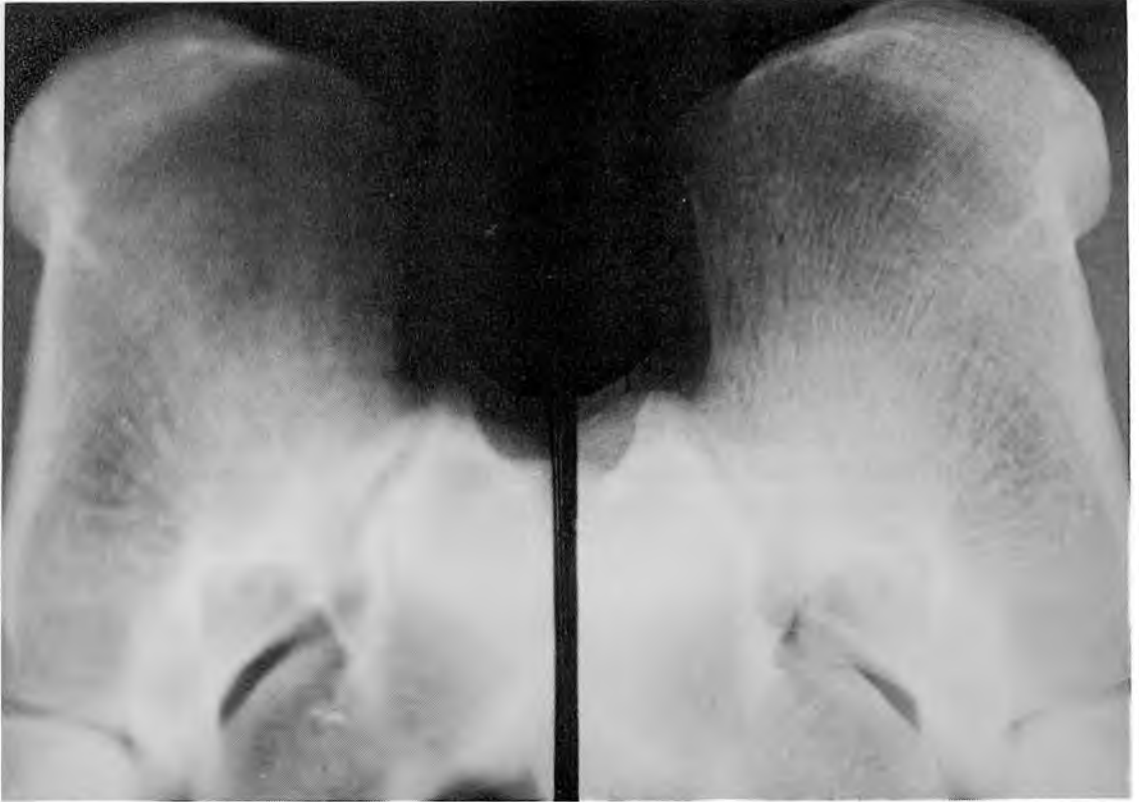


Fig. 22. Penumbra blur: roentgenogram of same part (heel of foot). Left. Large penumbra (2mm focal spot, 2X magnification). Right. Small penumbra (0.3mm focal spot, 2X magnification). (Courtesy of John Cullinan, F.A.S.R.T., Albert Einstein Medical Center, Philadelphia, Pa.)

small focus in the *same exposure*, required with increasing frequency in radiology.

The modern rotating anode tube combines high mA capacity *with* small focal spot size. This is achieved by using a disc-shaped anode which rotates rapidly. As a result, the electrons focus on a moving area of the target, so the heated area is effectively enlarged from a small rectangle to a nearly flat ribbon area near the edge of the disc. Of course, both the target area and filament are located off-center (Fig. 17). The anode is usually a cast molybdenum or tungsten disc mounted on a shaft attached to the armature of a motor enclosed within the evacuated glass envelope of the tube. Special metal bearings stably support the rotating assembly, which rotates about 3,300 revolutions a minute. (A new heavy-duty unit rotates 10,000 rpm.) An electric coil outside the tube provides a powerful ac magnetic field which drives the motor and anode assembly. It should be noted that the rotating anode motion is so beautifully controlled that there is negligible motion of the focal spot in space as the heavy anode rotates at high speed—a significant engineering accomplishment. Table 3(A)

compares the mA ratings for similar focal spot sizes of fixed and rotating anode tubes.

Tremendous amounts of heat are produced in some studies. For example, in one mammography technique, the anode is left at an orange heat following a single exposure! Such exposures are safe *individually* but can destroy the tube if they are delivered in too rapid succession. Modern x-ray tubes have been designed with very high heat dissipation capacities by using very massive anode structures of large diameter rotating at very high rates of speed. However, one must take the design ratings of all x-ray tubes very seriously or court costly replacement.

OTHER ANODE DAMAGE. The type of damage we have discussed up to this point is usually the product of exposures lasting only a few seconds at most. It results primarily from attempts to obtain very high x-ray intensities from x-ray tubes.

However, tubes can also be damaged at *low* mA values as well, if the current lasts too long. This is because substantial amounts of heat accumulate in the anode. This heat has ample time to spread out to parts remote from the tungsten anode, and damage occurs to parts made of other materials whose melting points are much lower than that of tungsten.

A dramatic example is damage to tubes by some inexperienced fluoroscopists. In fluoroscopy, currents are normally less than 5 mA. Due to its rather low mA and hence low operating temperature, the anode loses heat primarily by conduction to the bearings and the supporting structures; there is a limited rate of cooling.* If the fluoroscopist maintains too long exposures in order to stare at the fluoroscopic screen continuously, damage may occur to the tube in a matter of a few hours of operation. For example, a "Dynamax-40" x-ray tube may be operated at 100 pkV, 5mA for 30 minutes continuously, starting with the tube cold. Beyond this, it can be operated safely with on and off periods of ten minutes maximum. This is a generous design rating, adequate for any responsible operator, and exceeding the rating is difficult to justify. (Damage to the tube is only part of the story because the patient receives skin dosage at a substantial rate.)

In deep therapy tubes, general anode damage normally does not occur unless there is a cooling system failure. This eventuality is most probable in summer, when the water supply temperature is somewhat high and pressure lower than normal. For this reason, the temperature

* At high mA tube operation, the target temperature is very high, and much heat is removed by radiation to the housing. The rate of radiation is proportional to the fourth power of the absolute temperature, so radiation increases rapidly with temperature.

of the cooling system should always be checked before machine operation. Both local and general anode damage can occur if cooling is inadequate. Local damage can result from volatilization of tungsten and (infrequently) cracking of the target, with a consequent drop in x-ray output. On rare occasions, the target itself may be loosened in its copper block support as a result of local melting of the copper.

ANODE COOLING METHODS. The old non-shockproof tubes were frequently cooled by means of air-cooled fins, similar to devices used in refrigerators and air conditioners. These were screwed directly to the anode assembly and gave up their heat to the air. They are virtually never used by radiologists anymore because of electrical and radiation hazards.

Shockproof diagnostic tubes are sealed in oil, as previously mentioned. The larger tubes have large anodes capable of storing a great deal of heat. Fans may be mounted to cool the housing of such tubes, thereby permitting their continuous use in fluoroscopic work. Also, fans are sometimes helpful in preventing housing damage when heavy duty radiographic exposures are made repetitively.

No cooling is required in most very low kilovoltage x-ray therapy units (50 kV and lower). Treatment is of very short duration so that not too much heat accumulates. The time required for repositioning of the treatment fields is normally enough to permit safe dissipation of heat by the x-ray tube.

In more traditional superficial x-ray therapy (80 to 140 pkV), water cooling may be employed [Table 3(B)]. The housing interior is lined with cooling coils through which water flows. These cool the housing and oil and thereby indirectly cool the tube. Because the water is confined to the inside of these coils it cannot affect the insulating properties of the tube housing oil. In some other superficial therapy machines oil is pumped through the housing, then conveyed to heat transfer units where it is cooled by air or water before recirculation.

In x-ray machines operated at about 180 to 250 pkV, tubes are usually cooled in a more indirect manner, using a pumping system through which oil flows. The cool oil stream is directed against the material behind the target, from whence it flows around the outside of the x-ray tube. It then leaves the tube assembly through a hose at the cathode end.

Figure 23 shows the essentials of this type of system. The warm oil leaving the tube housing is cooled by a special cooler unit, usually remote from the tube, and then returned to the anode by a pump. The cooler unit utilizes ordinary tap water which is kept flowing while the machine is in operation. In many institutions the water is per-

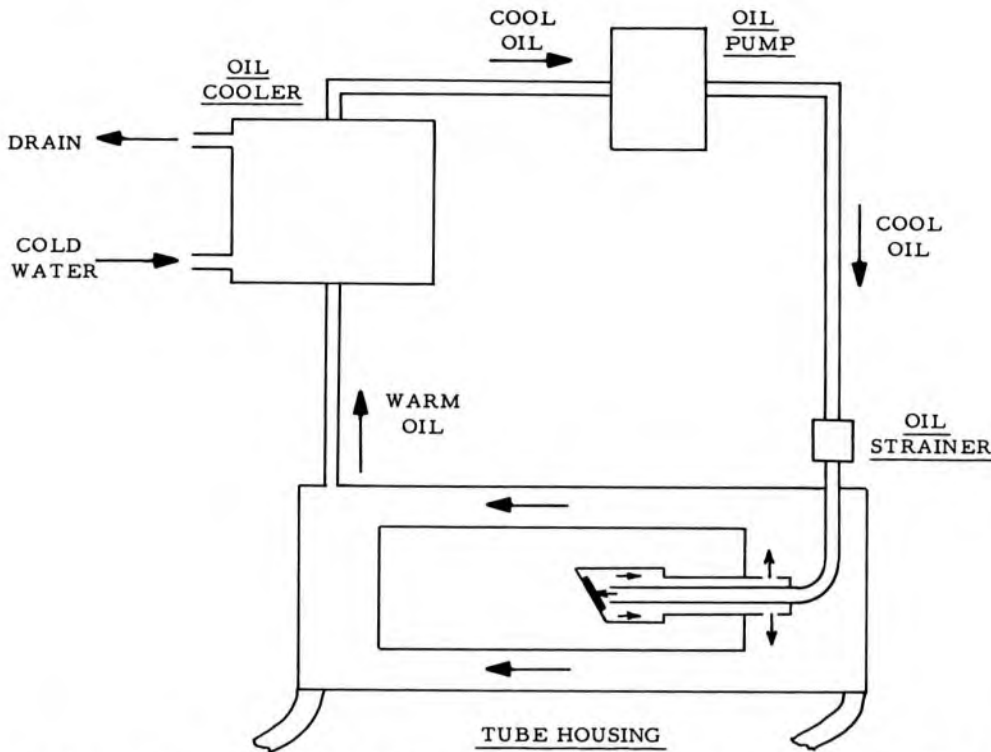


Fig. 23. Cooling system for deep therapy x-ray tubes (200-250 kV). Cool oil from the cooler flows through the copper anode which supports the tube target, then flows around the tube and out the other end of the tube housing and back to the water-cooled oil cooler. High voltage insulation must be maintained; therefore, the finest grade high dielectric strength oil is used and maintained scrupulously clean and dry.

mitted to flow continuously, even when the machines do not operate, to avoid the embarrassing consequences of a stuck solenoid valve. Usually a strainer is provided to remove dirt from the oil before it enters the tube housing. This strainer must be kept unclogged, or overheating can occur from reduction in oil flow.

A number of tubes operate with the anode structure at ground potential. Such tubes include those used in x-ray diffraction, self-rectified tubes used in deep therapy at 280 to 320 pkV, and most supervoltage x-ray tubes. Since the anode is already at ground potential in these tubes, it is most convenient to cool it directly by means of cold water flowing past the outer surface of the target.

TUBE RATING CHARTS. When fixed anode therapy tubes are used, their ratings are usually for continuous operation, so long as appropriate cooling is employed. The only admonition relates to the required cooling cycle and the maximum mA and kilovoltage operation permitted. These directions are usually stated on the specification sheet provided by the manufacturer and are relatively simple to follow, requiring no specific explanation here.

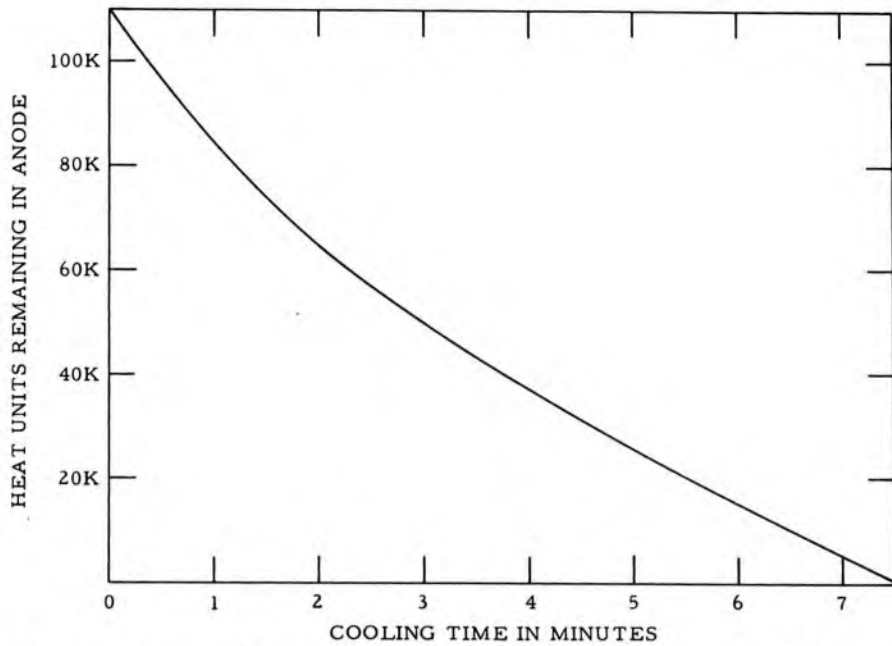
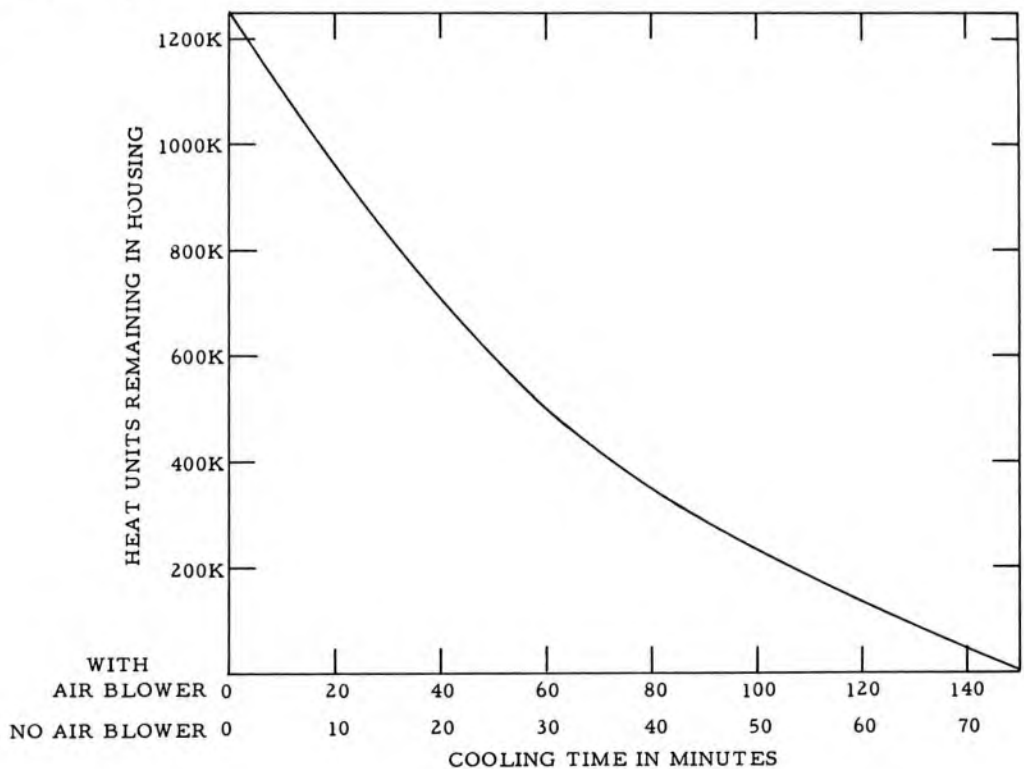


Fig. 24. Cooling curves for anode and housing of Dynamax-40 x-ray tube.¹⁸ Top. Tube anode. Maximum rating is only 110,000 heat units. The cooling rate, however, is quite rapid (30,000 h.u./min max), and cooling is virtually complete in 7.5 minutes. Bottom. Tube housing. Maximum rating is 1,250,000 heat units. However, even with an air circulator, cooling is relatively slow (30,000 h.u./minutes). *Without* a circulator, it is possible to damage the unit without anode damage! (Courtesy of Machlett Laboratories, Inc.⁸)



However, safe operation of rotating anode x-ray tubes for diagnosis is somewhat more difficult to assure. Corresponding to the two types of problems involved, there are two types of tube rating chart instructions.

1. *Permissible single exposure.* This type of chart tells the operator the maximum combination of tube milliamperage and exposure time permissible for different pkV values. Permissible values (manufacturer's data) vary with the following items:
 - a. *The focal spot.* At 100 pkV, a Dynamax-40 tube may be operated for one second at a maximum current of 140 mA, with a 1.0 mm focal spot. The corresponding current with a 2.0 mm focal spot is 275 mA.
 - b. *Supply voltage system.* For a Dynamax-40 tube with a 1 mm focal spot size, use of a self-rectified instead of full-wave machine at 100 pkV limits the permissible tube current to 67 mA rather than 190 mA.
2. *Long term cooling.* Figure 24 shows cooling curves for the anode and tube housing of the Dynamax-40 tube.

Note first that the *anode* can tolerate only 110,000 heat units,* as compared with 1,250,000 for the entire housing, because it is much smaller. If the 110,000 figure is exceeded, there is danger of anode damage.

The anode can lose heat only through the housing as a whole because the housing surrounds it. However, the ability of the housing to lose heat is also limited and can sometimes limit the exposure schedule. This ability is about doubled when an air circulator is used.

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* The heat units in an exposure = #mA × #pkV × #seconds of exposure (e.g., 2 sec exposure at 100 pkV, 100 mA yields 100 × 100 × 2 = 20,000 heat units).

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4

X-ray Attenuation

As shown in Chapter 3, the uses of x-rays result from their interaction with matter. In diagnosis, each part of the patient traversed by x-rays removes energy from the beam in its own characteristic way; the result is an array of x-ray shadows reflecting the body's structure. This shadow x-ray image is rendered visible by interaction with fluorescent screens or other detectors. In radiotherapy the x-ray beam must generally penetrate tissue to reach the tumor, and lose energy on the way. The therapeutic effect then results from absorption by the tumor of some of the remaining energy.

Attenuation is the removal of energy from a beam by matter it traverses. We shall consider attenuation in some detail because of its great radiologic importance. Our coverage includes a discussion of basic concepts, attenuation processes, and the nature and practical application of attenuation coefficients.

CONCEPTS AND DEFINITIONS

Several basic terms relating to attenuation must be carefully defined and explained to avoid confusion later.* In addition to these terms, this section also discusses the nature of transmission curves and attenuation coefficients.

Attenuation Terminology

INTENSITY, TRANSMISSION, AND ATTENUATION (Fig. 1). It will be recalled (Chap. 3) that the intensity of a beam is the rate of energy delivery perpendicular to a 1 cm^2 area [Fig. 1(Top)]. When the beam traverses matter it loses energy; thus, in Figure 1(Bottom) an ab-

* Fortunately these and other related terms and concepts have been reviewed recently by the International Commission on Radiological Units and Measurements, in Report 10a, 1962. Issued as N.B.S. H. 84. (See Appendix A.)

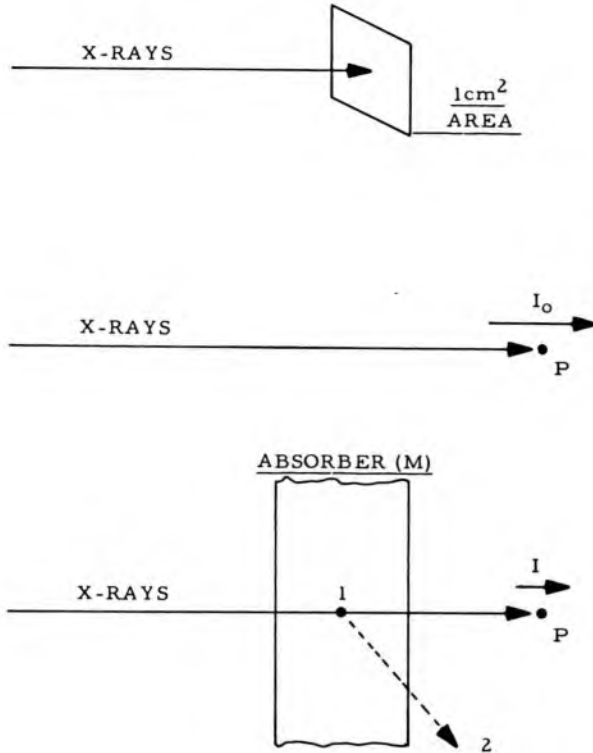


Fig. 1. Attenuation and transmission. Top. Intensity—energy per second normally incident on a 1 cm^2 area. Center. Intensity I_0 at location P—no absorber. Bottom. Intensity I at location P—slab of absorber M interposed. Intensity is reduced to I . Transmission is I/I_0 , always a fraction less than one. I is the transmitted intensity, and $(I_0 - I)$ was removed. The lost energy goes into the absorber (1. absorption attenuation) or is deflected out of the absorber (2. deflection attenuation). This consists of scattered photons in clinical work but can include other particles at very high energies.

sorber* reduces the beam intensity from I_0 to some smaller new value I . The ratio I/I_0 is the transmitted fraction, usually referred to as the transmission, (T) for short. T is always less than unity, since all materials absorb to some extent. The term attenuation is often used specifically to mean the fraction of I_0 which has been removed (A). One may write:

$$T = I/I_0 \quad [4-1(A)]$$

$$A = (I_0 - I)/I_0 = (1 - I/I_0) = (1 - T) \quad [4-1(B)]$$

As indicated in Figure 1 (Bottom), attenuation occurs in two basic ways. The first is by deposition of energy in the absorber, most generally by ionization. This is referred to as “absorption attenuation.”† The second is by deflection of energy out of the beam by atoms of the absorber; this is called “deflection attenuation.” In medical radiology deflection attenuation is generally by photon scatter.

It must be emphasized that these two terms describe only the over-all result of the various interactions. We shall see below that a

* The term “attenuator” is sometimes preferred, but “absorber” is still widely used and is retained here.

† The term “true absorption” has been used in the past, but “absorption” is now generally preferred.

single photon interaction with an atom may result in both absorption and deflection.

CONTRIBUTORY DEFLECTION. In practice x-ray beams may irradiate large volumes of material, and much energy is deflected in all directions. Under these conditions the transmission is found to depend to a considerable extent on the field size. Figure 2(Top) shows a beam of such small area (“narrow beam”) that substantially none of the deflected energy reaches B, a point directly beyond the absorber. Figure 2(Bottom) more nearly represents the practical situation; the field size is much greater (“broad beam”), and considerable deflected energy reaches B.

The radiation received at B thus consists of two parts: that transmitted directly and that deflected toward B by the absorber. Therefore, the location B receives more radiation from a broad than a narrow

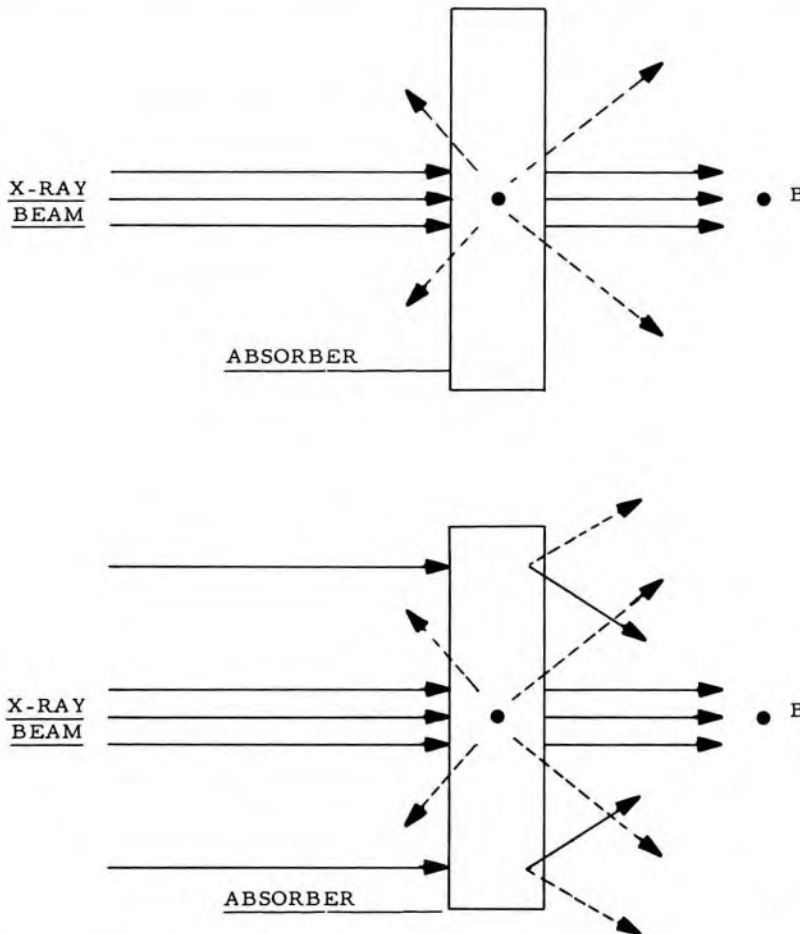


Fig. 2. Attenuation—narrow vs. broad beam. Top. Narrow beam: Radiation at B is almost exclusively directly transmitted. Attenuation is by absorption (dot) and deflection means (dotted arrows) only. Bottom. Broad beam: In addition to directly transmitted rays, B receives some *contributory scatter* from the absorber (solid diagonal lines). Hence, the transmitted intensity at B is greater than with a narrow beam.

beam of the same intensity, and the measured broad beam transmission is always greater. This fact is of great practical significance since the shielding effectiveness of any absorber is consequently always reduced by increasing field size.

Two applications serve to illustrate the importance of contributory scatter. Consider first a 2 MV x-ray therapy room in which the treatment beam is often aimed directly at the radiologist's desk. If one assumes only a narrow beam strikes the wall, a 24-inch concrete barrier is computed as adequate. However, the actual beam strikes a substantial wall area and produces a great deal of contributory scatter. As a result, 34 inches of concrete is actually required. Neglect of contributory scatter could result in the use of 10 inches too little concrete with transmitted intensities about 12 times higher than predicted. Consequently, radiation protection calculations always employ broad beam data (Chap. 15).

As a second illustration, Figure 3 illustrates the contributory scatter in 200 kV x-ray therapy. (In clinical work, virtually all con-

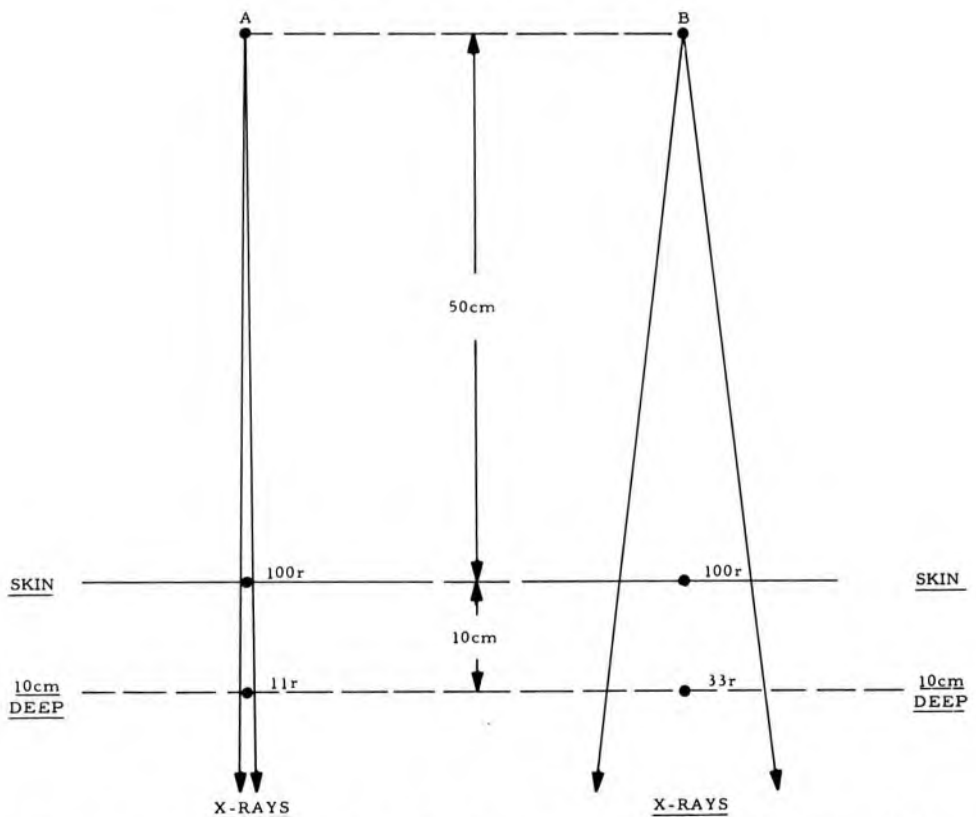


Fig. 3. Contributory scatter in 200 kv x-ray therapy. A. Tiny field—without much contributory scatter, only 11 r is delivered to a tumor at 10 cm depth, for 100 r to skin. B. 10 x 12 cm field—with more typical contributory scatter, 33 r is delivered. Thus, in typical orthovoltage therapy, about two-thirds of the dose may be delivered by contributory scatter radiation.

tributory deflection energy consists of scattered photons, so from here on we shall use this shorter term.) Figure 3(A) shows the dosage, using a very tiny field (about 1 cm²), delivered at 10 cm depth to a patient; in (B) everything is kept the same, but a larger field (10 × 12 cm) is used. The larger field (broad beam) delivers three times more dose at depth than the tiny one for the same skin dose (100 r). It is evident that contributory scatter is responsible for most of the tumor radiation in orthovoltage therapy, since substantial fields are usually employed to encompass tumor-bearing areas.

Geometric Effects—Beam Divergence

So far we have considered the effect of attenuation only, confining our attention in all cases to one location at a time (points P, B, and the tumor at 10 cm depth in the three figures). Any observed loss in beam intensity therefore results from interposition of material in the beam. We now consider the effect of varying distance from the radiation source on beam intensity.

REDUCTION OF BEAM INTENSITY BY BEAM DIVERGENCE. By definition all radiation spreads out in all directions from its source. X-ray beams therefore lose intensity *even when there is no absorber present*, as a result of beam divergence alone. Figure 4 shows x-rays emanating

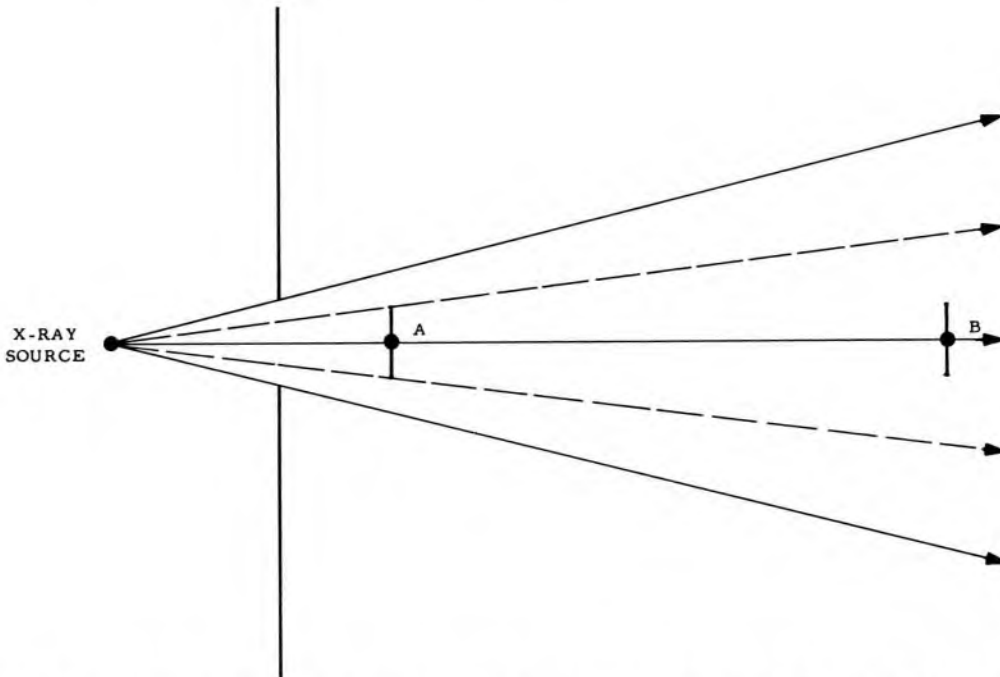


Fig. 4. The effect of x-ray divergence: Some of the radiation hitting an object at A will miss it at location B (dashed lines). The *intensity* at B is hence lower than that at A because of beam divergence.

from a source. At point A, an object of given size is struck by all the photons between the two dashed lines. At the more remote location, three times farther from the source, only those photons originally confined to an angle roughly one third as large strike B; those beyond this angle miss B completely. In general, the more remote object is struck by a smaller fraction of the available photons: simple geometry indicates this fraction should be inversely proportional to the square of the distance involved. Thus, if B is three times as far from the source as A, the intensity is only one ninth as great. This is simply a divergence, or geometric phenomenon. Divergence operates independently of attenuation, and the observed intensity relationships in practice are most generally due to the combined effects of attenuation and beam divergence.

EXAMPLES. We shall give three examples to illustrate beam divergence or geometric effects.

First, suppose one desires to perform an x-ray diagnostic study at a 30-inch target-film distance instead of the 20-inch usually used. Since the distance from the source has been increased 1.5 times, the beam intensity is reduced at the film 1.5×1.5 , or 2.25 times, and hence the exposure must be correspondingly increased to adequately expose the film.

Similar relationships are noted in radiation therapy. Consider a problem in orthovoltage therapy. The radiotherapist wishes to increase his treatment distance (tube target to patient's skin) from 50 to 70 cm. This increases the tumor dose for a given skin dose, a desirable result as we shall show below. Unfortunately the increase in distance by a factor of $(7/5)$ decreases the beam intensity by $(7/5)^2$, or about two times. The required treatment time is correspondingly nearly doubled. Since useful treatment beams at 200 kV are of rather low intensity, the required new treatment time may be excessively great.

Finally, it is instructive to discuss briefly a somewhat more complex but very important application of this principle (Fig. 5). Consider two x-ray machines used to fluoroscope the same 8-inch thick patient, who will be placed between A and B in each unit. For the doctor to see the image, a minimum x-ray intensity must reach the screen in both cases. Since the patient attenuation is constant, only the divergence attenuation is different in the two situations.

How important is this difference? In Figure 5(Left) divergence alone reduces the intensity as the beam travels from A to B by a factor of $(14/6)^2$ or about 5 times. At right the reduction is only by a factor of $(26/18)^2$ or about 2 times. Hence, the divergence attenuation at left is 2.5 times that at right. As a result, for the same brightness on

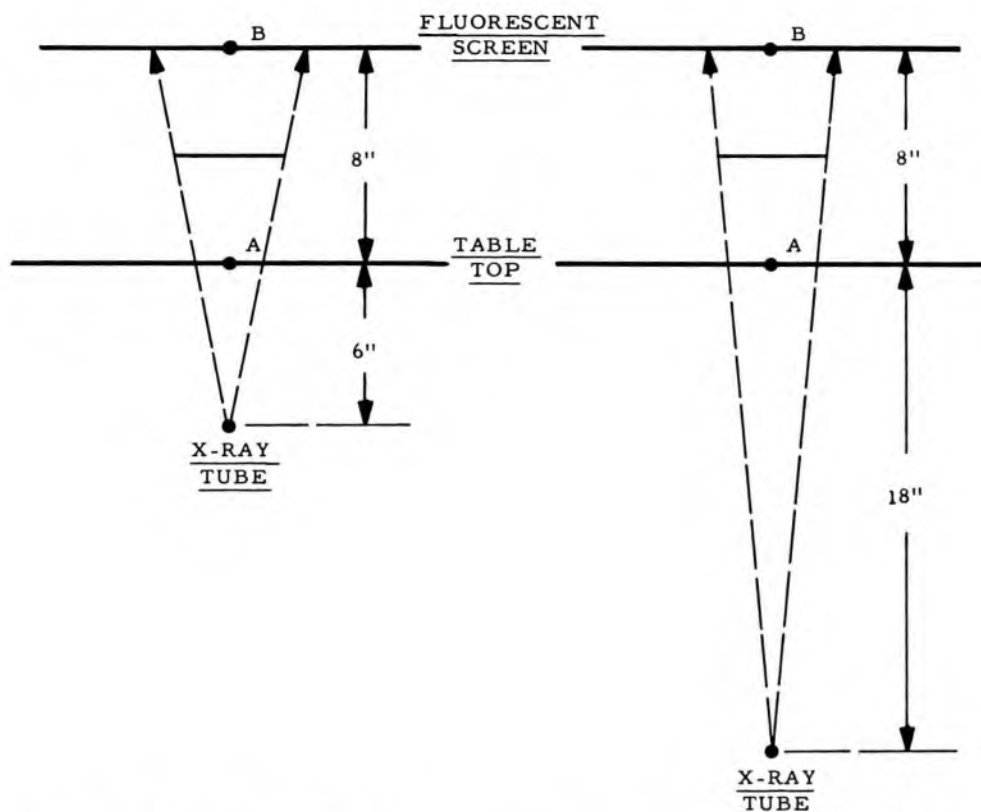


Fig. 5. Beam divergence in fluoroscopy. Left. Tube too close to table: Intensity at the screen can be only one-fifth that at A even without attenuation by patient. Right. Approved distance: Corresponding factor is one-half versus one-fifth. X-ray attenuation is substantially the same in both cases (same parts traversed by beam). Hence, the patient's skin near table receives 2.5 times greater dose in A than B, for same brightness on the screen.

both screens the required patient's skin dose at A is 2.5 times greater at left than at right!

This type of argument has resulted in the setting of a 12 inch minimum target-panel distance in most State Radiation Protection codes. (See Chap. 14.)

Transmission Curves

As previously shown, the penetrating ability of x-ray beams is of great practical interest. This ability is usually evaluated in terms of the beams' transmission through particular materials. Such data also provide the basis for discussing attenuation coefficients.

Narrow beam transmission is most uniquely characteristic of the spectrum and absorber combination because broad beam data vary with field size. We shall hence consider primarily narrow beam data,

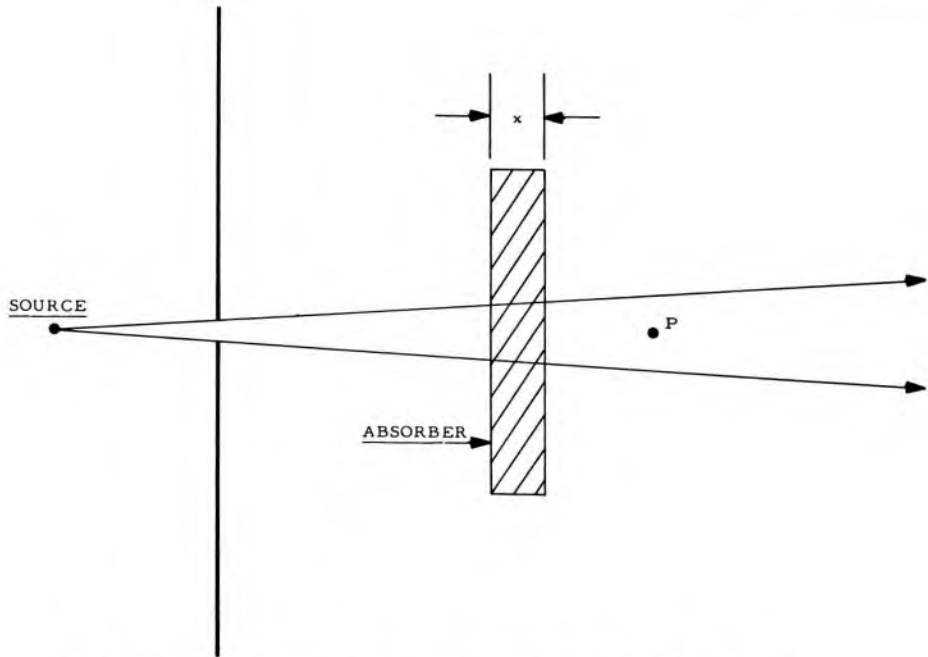


Fig. 6. Definition of “transmission” (see text).

first for monochromatic x-ray beams and then for actual bremsstrahlung beams.

TRANSMISSION VERSUS ABSORBER THICKNESS. Consider an experimental arrangement for measuring x-ray transmission (Fig. 6). One first measures the beam intensity at P without any absorber (I_0), then with increasing thickness of absorber (I). The transmission (T) is given by [4-1(A)] above as I/I_0 . Recall that divergence is not involved in this definition because intensity is measured at only one location throughout this discussion.

Our common experience would indicate that as absorber thickness increases so does attenuation, resulting in decreased transmission. This is observed experimentally. When careful measurements are carried out using monochromatic x-rays, the following relationship results:

$$T = e^{-\mu x} \quad (4-2)$$

where x = absorber thickness

μ = a constant characteristic of a particular photon energy and absorber

$T = I/I_0$ = narrow beam transmission

This is an example of a decaying exponential function, which is one of the most important relationships in radiation physics. It occurs

in both attenuation of x-rays and some other ionizing radiation and in analysis of radioactive decay (Chap. 9).

GRAPHIC REPRESENTATION OF T VERSUS x. In dealing with exponential functions it is often more convenient to use graphic rather than analytic representations. Figure 7 shows three ways in which this type of function can be graphed. The first [Fig. 7(Left)] uses ordinary graph paper. Initially, the resulting curve falls rapidly and then levels off, approaching the horizontal axis closer and closer as it goes farther and farther out, but never quite reaching it (asymptotically). This simple graph is very hard to use beyond the initial relatively straight portion of the curve. Since straight line graphs are much easier to use accurately, the second way of plotting the graph [Fig. 7(Center)] is preferable. The curve is a straight line because the logarithm of the transmission ($\log T$) instead of T itself is used for the ordinate. While this is easier to graph and to read, it is inconvenient to take logarithms. It is far easier to do this automatically by using semilog graph paper. Figure 7(Right) shows the same relationship plotted with this graphical tool. Ordinarily, reasonably accurate x-ray transmission or radioactive decay curves can be obtained using semilog graph paper.

VARIATION OF TRANSMISSION WITH PHOTON ENERGY AND ABSORBER. The slope of the graph at right depends on the photon energy and type of absorber used. Another way of saying the same thing is that

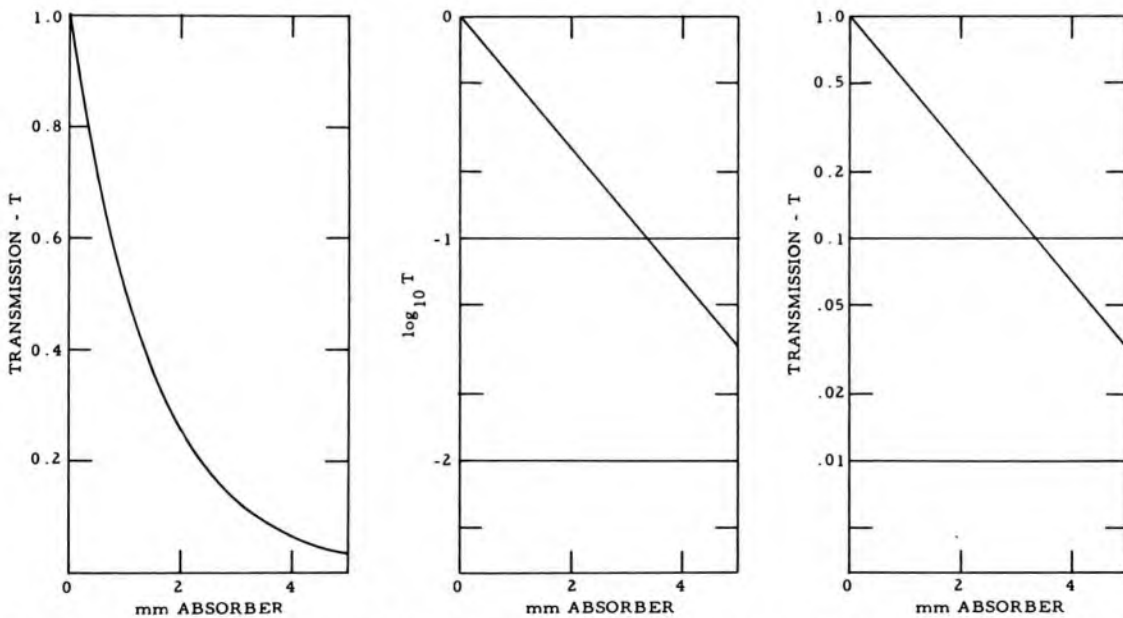


Fig. 7. Three ways of plotting a decaying exponential relationship. Left. Ordinary graph paper: T vs. x plot yields curve. Center. Same paper: $\log T$ vs. x plot yields straight line. Right. Semilog graph paper: T vs. x plot yields straight line.

μ in equation (4-2) depends on the photon energy, the absorber density, and atomic number.

High energy photons are generally more penetrating than those of low energy. This is illustrated in Figure 8(Top), which shows the transmission curves for 50 and 20 keV photons traversing aluminum. The 20 keV energy is typical of very soft photons, the 50 keV of harder or more penetrating photons in a diagnostic x-ray beam. Note that 20 keV photons are reduced 90 percent by traversing about 3 mm of aluminum, while 50 keV photons are reduced far less. To reduce them by 90 percent requires a full 10.5 mm of aluminum.

Figure 8 (Bottom) compares transmission curves for 50 keV photons in copper and aluminum. Here the atomic number is changed rather than photon energy. Evidently copper is a much more effective absorber of 50 keV photons than aluminum, far more than explainable by its being about three times denser.

ATTENUATION COEFFICIENT μ . The attenuation coefficient μ is a constant determined by the particular combination of absorber material and photon energy. It is a measure of the ability of an absorber to attenuate photons. Equation (4-2) can be rearranged to express μ explicitly:

$$\mu = -(1/x) \log_e T \quad (4-2')$$

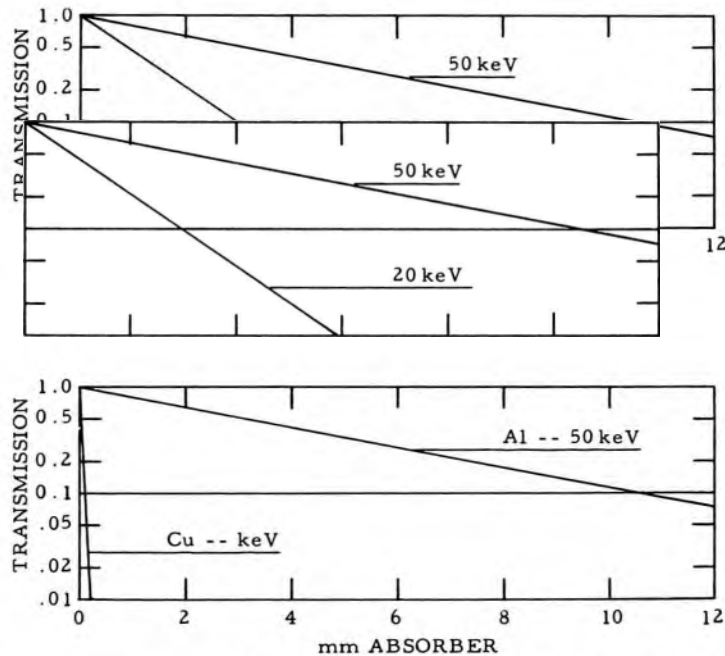


Fig. 8. X-ray transmission depends on both photon energy and absorber. Top. Photon energy: curves for 50 and 20 keV photons traversing aluminum. Bottom. Absorber material: curves for 50 keV photons traversing aluminum and copper.

This relationship effectively defines μ and can be used to compute μ from measured values of T and x. For example, 0.1 mm of copper absorber yields a transmission of 0.960 for 100 keV photons. Substituting in (4-2') yields:

$$\mu = -(1/.01\text{cm}) \log_e(0.960) = (1/.01\text{cm})(-.0410) = 4.10/\text{cm}$$

X-ray attenuation depends directly on the number of electrons encountered by 1 cm² of the beam in traversing the absorber. If the density (g/cm³) changes, so will the attenuation (e.g., air versus compressed air). Hence, the attenuating ability per gram is often desired. This is obtained by dividing μ by the absorber density and is called the mass attenuation coefficient, with units of cm²/g. (See below.)

In the above example, the density of copper is 8.93 g/cm³; hence

$$\mu/\rho = \frac{4.10/\text{cm}}{8.93 \text{ g/cm}^3} = 0.459 \text{ cm}^2/\text{g} \quad (4-3)$$

ACTUAL TRANSMISSION CURVES. The curves considered up to now were all for monochromatic beams only. However, actual x-ray beams almost always have a continuous spectrum, with photons of a wide range of energies (Fig. 12 in Chap. 3). These are continuously distributed between a maximum energy determined by the pkV and a minimum determined by the total filtration.

Figure 9 illustrates the general nature of the transmission curve for actual x-ray beams (middle curve). Curves A and B correspond to the transmission curves one would obtain for monochromatic beams having the maximum and minimum photon energies in this continuous spectrum. The actual beam contains photons whose energies lie between these two values. Therefore, its curve should always lie between these two limiting curves. However, it cannot be a simple straight line like the others, which correspond to monochromatic rays, because the actual beam changes as it penetrates the absorber. Initially, the beam contains substantial amounts of low energy photons which are disproportionately attenuated by relatively small thickness of absorber. By virtue of the more selective removal of these photons, the beam is made both harder and more homogeneous. The net effect is that as absorber is added, the curve drops rapidly at first, then levels off to a reduced slope and falls more slowly. When thick absorbers are used the beam becomes relatively hard, and its slope tends to approximate that of the highest keV present.

X-RAY BEAM QUALITY. Before leaving the subject of x-ray transmission curves, let us return briefly to the subject of specifying x-ray

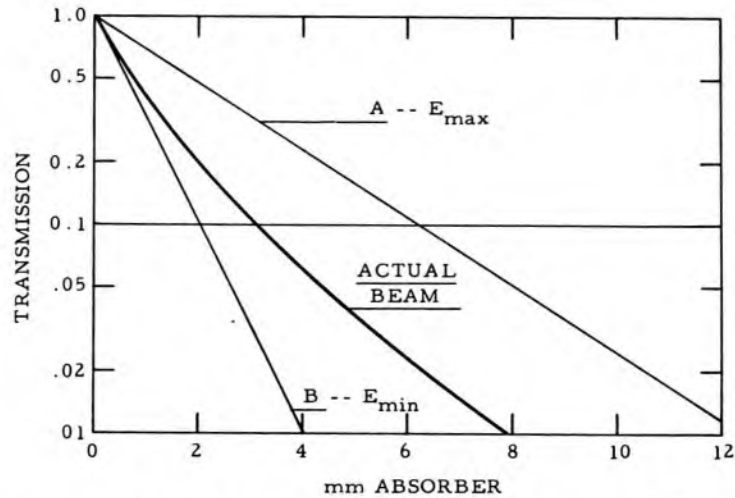


Fig. 9. Actual transmission curves. The beam's spectrum lies between the limiting values E_{\max} and E_{\min} (Fig. 12, Chap. 3, p. 92.) Hence, its transmission curve lies between the curves for these photon energies. The slope is initially great, approximating that of the E_{\min} curve (B) because most of the initial attenuation involves lower energy photons. At higher absorber thicknesses, the curve slope approximates that of E_{\max} (A) because low energy photons have been substantially removed.

beam quality. Because x-ray spectra are difficult to measure in practice, the half-value layer (HVL) is usually employed to specify the x-ray beam. This is the thickness of specified material which attenuates the given narrow beam 50 percent. For radiation therapy it would be useful to use tissue as a reference material. In diagnosis and therapy, it is convenient to use metals as substitutes for tissue, to obtain convenient HVL values in the range of millimeters which can be readily measured reproducibly. In general, aluminum is used for HVL measurements for beams generated at 130 pkV and less. For orthovoltage x-ray therapy, copper is employed; this covers the range of about 130 pkV up to several hundred pkV. For supervoltage, either copper or lead is preferred for HVL measurements. Table 1 presents some typical HVL values for x-ray beams useful in clinical radiology.

Several other indices of x-ray beam quality have been employed in the past, of which the "effective energy" is perhaps one of the most common. This is defined as the energy of a monochromatic beam which has the same HVL as the continuous spectrum beam in question. Of course, this effective energy also corresponds to an effective wavelength, a term which has also been employed. Both HVL and effective energy designation of x-ray beam quality have their limitations because a monochromatic beam can only roughly approximate a bremsstrahlung beam in its x-ray transmission curves. Different combinations of pkV and filtration can yield the same HVL values *in metal*, and yet the penetration of their beams *in tissue* may be entirely dif-

TABLE 1. TYPICAL HVL VALUES FOR COMMON FACTORS

pkV	Wave Form	Filter	Typical HVL	Corresponding Effective keV	Application
80	Rectified	3 mm Al	2.5 mm Al	40	Diagnosis
100	Rectified	Inherent	1.0 mm Al	35	Therapy—superficial lesion
100	Rectified	1 mm Al	2.0 mm Al	40	Same—slightly deeper
140	Villard	$\frac{1}{4}$ Cu + 1 Al	0.5 mm Cu	75	Therapy—intermediate
200	Villard	Thoraesus II	1.6 mm Cu	100	Deep therapy
250	Constant	Thoraesus III	3.0 mm Cu	135	Deep therapy
2,000	Constant	0.1" Au	8.0 mm Pb	900	Supervoltage therapy
6,000	Pulsed	*	12.7 mm Pb	1,750	Supervoltage therapy

* Filtration varies with type of machine and manufacturer, depending on target materials and compensating filter design.

ferent. However, in practice, reasonably similar operating kilovoltages are employed to obtain given HVL values, so the HVL does serve a useful function. When one wishes to describe a particular x-ray beam, not only the HVL, but also the pkV, filter and wave form should be specified to guide another person in duplicating the conditions of an original irradiation.

ATTENUATION PROCESSES—DESCRIPTION

We have so far considered what happens to the beam intensity during x-ray attenuation but not how the attenuation takes place. We now consider some of the processes involved. All of these processes but one (simple scatter) result in primary ionization of the medium.

Table 2 shows five of the mechanisms by which photons interact with atoms; also given is the energy range in which each is most important in water or soft tissue. Only the first four occur at usual operating kilovoltages; even the fourth mechanism occurs to a significant extent only in irradiation by betatrons and linear accelerators. In our discussion, we shall give a symbolic description of the reaction and account for the fate of all particles and energy.

Note in Table 2 that the mechanisms are indicated in order, with those most important at low energies at the top. We follow this sequence in our discussion below.

Simple Scatter Attenuation

This is also referred to as Thomson, coherent, and unmodified scatter. Figure 10 shows a simplified picture of what occurs. A photon, indicated by the wave with the arrow beneath it, interacts with an atom and is simply deflected in direction. The atom remains intact, as does the photon, to all intents and purposes. The atom has simply provided a means to redirect the photon impinging on it but has acquired no significant energy, so there is no *absorption*. However, the photon has been deflected from the main beam, so there is *deflection attenuation* because an object placed beyond the absorber would

TABLE 2. FIVE X-RAY ATTENUATION MECHANISMS—
IMPORTANCE IN SOFT TISSUE

Mechanism	Energy Range of Maximum Importance to Soft Tissue	Attenuation	Remarks
1. Simple scatter (Thomson, coherent, unmodified)	1 keV to 200 keV		Less than 5 percent of total, generally. Max. of 9 percent of total at 30 keV.
2. Photoelectric	1 keV to 29 keV		Decreases very rapidly with energy: at 15 keV, accounts for 80 percent; at 60 keV, 10 percent of total attenuation.*
3. Compton scatter (Incoherent, modified)	29 keV to 20 MeV		Dominant mechanism in almost entire clinically useful range. Changes very slowly with keV.
4. Pair and triplet production	20 MeV and higher		Starts at 1.022 MeV. Only 10 percent of total at 4 MeV. Significant in beta-tron but not cobalt-60 or 2 MV supervoltage work.
5. Photodisintegration	Peaks at 24 MeV		Zero below 15 MeV in tissue. Negligible biological contribution.

* In contrast, photoelectric effect is dominant in transferring energy to the medium up to about 57 keV. (See Table 9.)

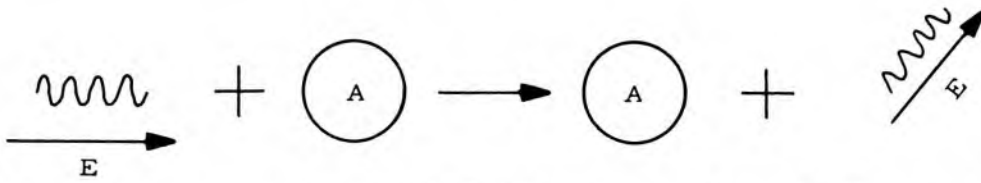


Fig. 10. Simple scatter (unmodified, Thomson, coherent). Here, a photon is simply altered in direction with no significant change in its energy or communication of energy to the atom deflecting it.

not be struck by the original photon. Since the photon energy is not altered significantly the process is called “unmodified scatter.” It is similar to that occurring to lower energy photons, such as those of radio and light waves.

Simple scatter is of secondary importance in most x-ray applications. For photons of energy 10 keV and greater, it contributes far less to attenuation than the photoelectric effect and the other type of scattering process, Compton scatter.

Photoelectric Interactions

Figure 11 describes this process schematically. The photon of energy E interacts with the atom as a whole and disappears completely in the process. The atom is left ionized, most often by loss of a

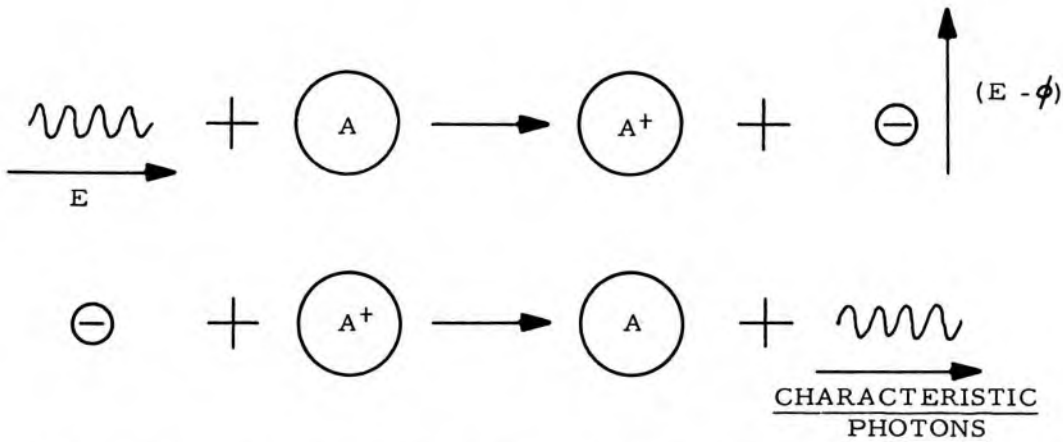


Fig. 11. Photoelectric interaction. Top. The ionization itself. The incoming photon gives up *all* its energy and hence disappears. Its energy E is used first to remove an orbital electron by supplying the required ϕ worth of energy. The remaining $(E-\phi)$ is then given to the electron as kinetic energy. The atom is left ionized, having lost an electron. Bottom. The aftermath. The ionized atom is restored by acquiring a free electron. It gives up the ϕ 's worth of energy as one or more characteristic photons, or in producing Auger electrons (see text). The original escaping electron ionizes and excites atoms in its path until slowed down to thermal speeds, normally some distance from the original atom.

K-orbit electron. The photon energy E is divided into two parts. The first is used to ionize the atom (ϕ keV in the figure). The second appears as energy of motion of the electron ($E - \phi$). (The released electron is called a "photoelectron.") ϕ is the binding energy of the orbit from whence the photoelectron came and is characteristic of the atom and orbit involved.

The released photoelectron generally has considerable energy and ionizes atoms in its path until this energy is dissipated. Thus, $(E - \phi)$ of the original photon energy is absorbed rather than deflected in the photoelectric interaction.

RESTORATION OF THE ATOM. What of the atom originally ionized? It is soon restored to neutrality by the capture of a free electron, and the energy ϕ originally deposited during the atom's ionization is released. Most frequently, the K orbit is restored by a nearby L-orbit electron, which is closest; the new L-orbit vacancy is, in turn, filled by more remote electrons. Several photons may thus result; however, their total energy must total ϕ , the energy available for their formation.

The energy ϕ may also be released by an alternative process: the Auger effect. In this process, the energy ϕ ionizes an orbit of lower binding energy than the original, releasing an "Auger electron." Just as photoelectrons, Auger electrons give up their energy by ionization. The Auger effect competes strongly with characteristic photon release in low Z absorbers like tissue. For example, it constitutes 95 percent or more of the total energy release during atom restoration, for elements up to sulfur in the Periodic Table, and 88 percent for calcium. Characteristic radiation becomes more important for high Z absorbers, reaching 86, 94, and 96 percent of the total for iodine, tungsten, and uranium, respectively.

ENERGY ABSORPTION. In biologic materials photoelectric attenuation is virtually pure absorption. This is mainly because ϕ values are generally 1.5 keV and less, and the incoming photon energy is much greater. In addition, any characteristic low energy photons are readily absorbed by tissue before they can escape the body. Of course, there is relatively little characteristic radiation produced in tissue, since the Auger effect is dominant.

Photoelectric attenuation is dominantly absorption even in high Z materials because characteristic photons tend to be absorbed in the absorber before they can escape. However, in some situations escaping photons may require specific attention, as in filter design, as we shall show below.

LIKELIHOOD OF PHOTOELECTRIC INTERACTIONS. Photoelectric attenuation is a resonance affair which is most likely to occur at photon

energies just above the binding energies of the involved orbits. In general, the amount of attenuation diminishes rapidly as the photon energy increases beyond the ϕ values. In addition, it rises rapidly with absorber Z . For example, the photoelectric attenuation by electrons of bone mineral is about 11 times greater than that by surrounding soft tissue. Consequently, bone absorbs more in the low keV region, where photoelectric effect is the dominant attenuation mechanism. At higher photon energies, where the photoelectric effect is negligible, this distinction substantially disappears, a fact of considerable importance in chest radiography. (See Chap. 8.)

Compton Scatter Attenuation

Compton scatter is also referred to as *incoherent* and *modified* scatter, in contrast to Thomson scatter which is *coherent* and *unmodified*. Figure 12 shows the events involved in a Compton interaction. The incoming photon interacts with an electron, which is generally in the outer orbit of the atom. The reaction resembles that between two billiard balls. The electron is always driven in a forward direction and emerges at an acute angle with respect to the original photon direction (angle θ in the figure). The photon may be deflected through any angle from 0 to 180° . The electron is referred to as a "recoil electron" because it has undergone something akin to a mechanical collision. The photon has less energy after scattering because it shared some

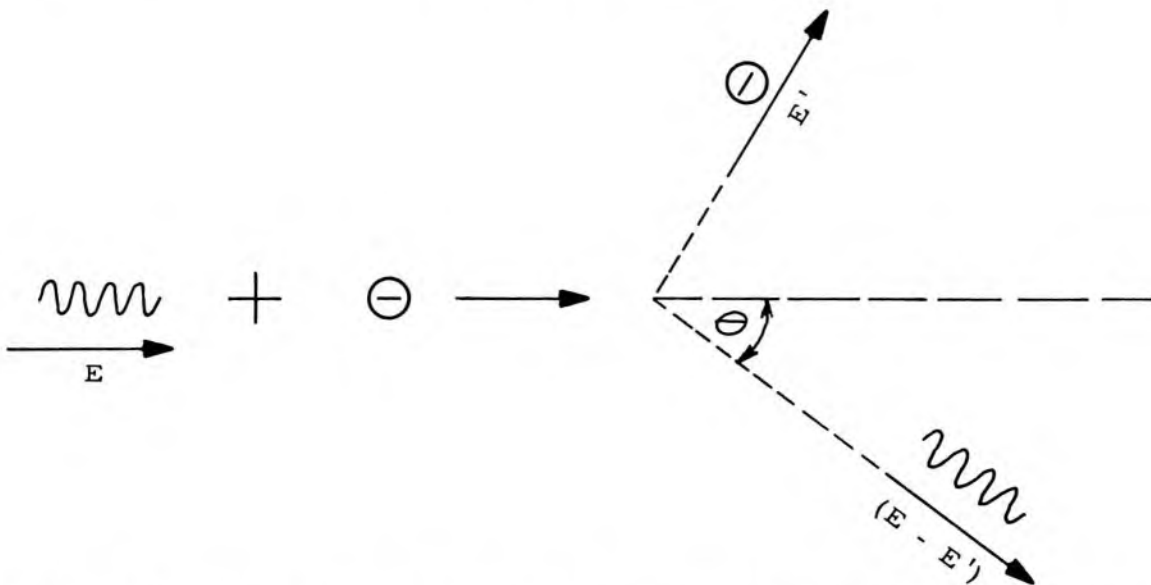


Fig. 12. Compton scatter (modified, incoherent scatter). The incoming photon interacts with an electron, normally one from an outer orbit. The electron is driven in a generally forward direction and the photon deflected. The photon gives up energy to the electron. It is therefore reduced in energy by this same amount and increased in wavelength.

with the recoil; its wavelength is therefore correspondingly increased, and the photon appropriately said to be "modified." In essence, part of the energy of the incoming photon has been given to the recoil electron. However, unlike the photoelectric event, much of the photon's initial energy is retained, and the photon remains, although with less energy.

The atom from which the recoil electron came is of course left ionized. However, only a few eV is required to ionize the outer orbits of atoms; this is negligible compared with the thousands of eV normally communicated to the recoil electrons in medical radiology. Any small amount of characteristic radiation released when the atom is restored is in the light range and is generally absorbed within the irradiated material.

The recoil electron has considerable energy and ionizes many atoms in its path until this energy is dissipated. The recoil's energy hence constitutes absorption rather than deflection attenuation. Of course, if the modified photon escapes the absorber, its energy represents deflection attenuation.

SCATTERED PHOTON DIRECTIONS. Very low energy photons (below 10 keV) tend to be deflected primarily by simple scatter. In this process photons are scattered in backward and forward directions with essentially equal frequency, with considerable scatter in a lateral direction as well. Higher energy photons are scattered preferentially in a forward direction; this effect becomes increasingly great as photon energy is increased. For example, 50 keV photons are scattered forward about 40 percent more frequently than backward; 500 keV photons, 5 times; and 2 MeV photons, over 15 times more frequently. Lateral scatter is also reduced at higher photon energy, although to a less degree than back scatter.

Thus, higher energy or "supervoltage" x-rays tend to be scattered forward rather than backward or to the side. As we shall see in Chapter 8, this fact contributes to the advantages of supervoltage radiation in radiotherapy of deep-seated lesions.

WAVELENGTH AND ENERGY OF MODIFIED PHOTONS. The change of wavelength (\AA) suffered by the photon in Figure 12 is given by a simple formula:

$$\Delta\lambda = .024 (1 - \cos\theta) \quad (4-4)$$

This change is independent of the original photon energy, a rather remarkable result. It is instructive to compute the new wavelength and photon energy of Compton scattered photons of different initial energies for various deflection angles (Table 3). Some rather

TABLE 3. ENERGIES OF COMPTON SCATTERED VS. INCIDENT PHOTONS, FOR VARIOUS DEFLECTION ANGLES (θ)

Initial keV	Scattered Photon Energy in keV for:			Average Energy* of:	
	$\theta = 30^\circ$	$\theta = 90^\circ$	$\theta = 180^\circ$	Scattered Photon	Recoil Electron
10	9.9+	9.8	9.6	9.85	0.15
20	19.9+	19.25	18.55	19.38	0.62
50	49+	45.5	41.8	46.0	4.0
100	99+	83.6	71.8	86.0	14.2
200	190+	143.5	114.2	156.5	43.5
500	442	252.5	169.0	335	165
1,000	790	338.0	203.0	560	440
2,000	1,310	404.0	726.0	900	1,100
Extremely great	3,830	514.0	257.0	----	-----

* All angles of scatter included.^{1,2}

important practical consequences are evident from the resulting data; these are summarized briefly below.

1. In both clinical diagnosis and in orthovoltage radiation therapy (10 to 200 keV), there is little practical difference in quality between the scattered and incident x-rays.
2. However, in the supervoltage and cobalt-60 therapy range (500 to 2,000 keV), the beam is greatly softened by Compton scatter. For deflection angles of 90 degrees and greater (i.e., photons deflected through right angles or scattered back toward their source) the energy of a scattered photon is always less than 514 keV, regardless of how high its initial energy may be! This is a very important result in radiation protection barrier design because it permits the use of thinner shields than would otherwise be required. However, for deflection angles less than 90 degrees, the beam is softened far less, and tends to be more similar to the original beam.
3. Recoil electrons tend to have very low energies compared with photoelectrons produced by photons of the same energy.* This is because a photoelectron has virtually *all* the energy of the incident photon while the recoil electron usually has only a *small part* of this energy.

* Up to 100 or 200 keV initial photon energy. Beyond 200 keV photoelectric absorption is of relatively little importance in most tissue materials.

Pair Production

Figure 13 (Top) shows what happens in pair production. In this interaction a very high energy photon interacts with the nucleus of an atom. The atom is essentially unchanged in the process and serves a purely catalytic function. As a result of the interaction, two new particles are produced, seemingly out of nothing! The photon is completely destroyed, being *transformed* into two new particles, with energy left over appearing as particle kinetic energy. One particle is an ordinary electron, like those in orbits of atoms. The other is novel and exotic: a *positively charged electron*, called a “positron.” Since momentum is conserved, these particles tend to travel in roughly opposite directions following their production. They are called a “pair” because they have identical masses and magnitude of charge even though one is plus and the other minus. Since they were obviously produced, the process is referred to as *pair production*.

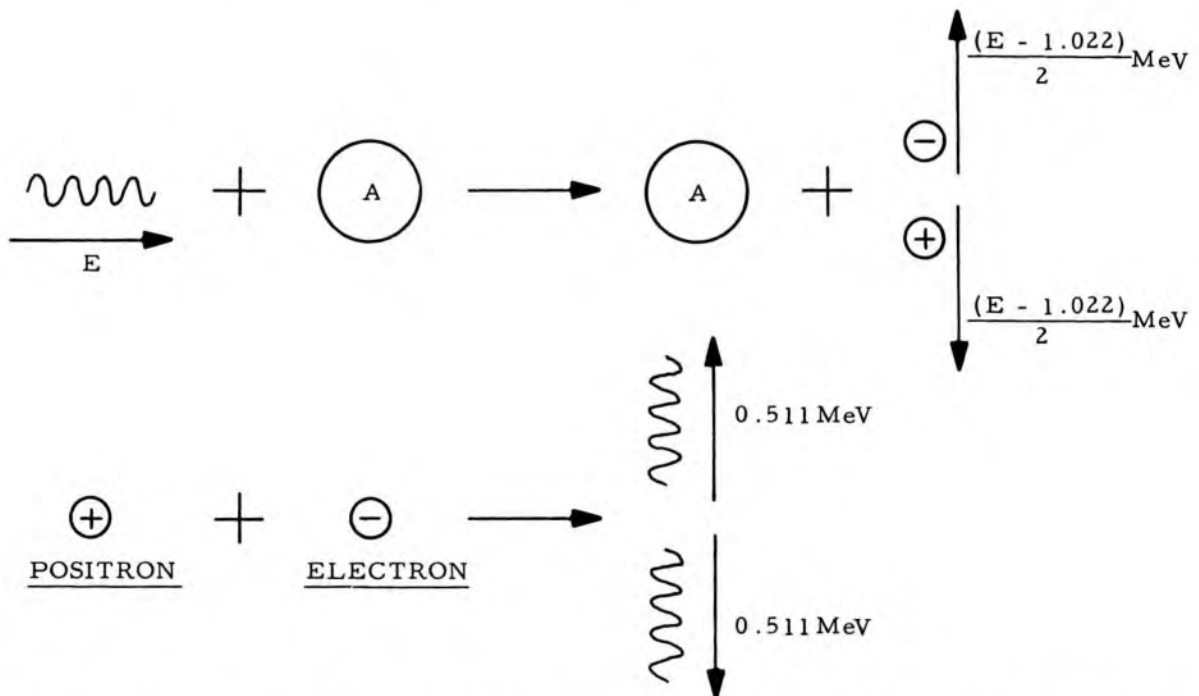


Fig. 13. Pair production. Top. The process itself. Note the photon disappears completely. Part of its energy appears in material form as an electron pair (an electron and a positron). The remainder is divided unequally between these particles and appears as energy of motion. Bottom. Annihilation. Both the ordinary and the positive electron give up any energy of motion in collisions with atoms they encounter, ionizing and exciting them. The negative electron ends up as a relatively slow free electron, like others already present. As it slows down, the positron develops a greater and greater probability of entering a fatal union with an ordinary electron: annihilation. This finally destroys both participants, generally producing two 0.511 MeV photons as shown.

The two new particles were produced from the energy of the original photon. This is another example of the miracle, discovered only sixty years ago, of the transformation of energy into matter, and in other situations matter into energy. (In the atom and hydrogen bombs, matter is transformed into energy.)

The arithmetic is relatively simple. Since an electron or positron weighs 9.11×10^{-28} gram at rest, the pair weighs a total of 18.22×10^{-28} gram. This corresponds to 1.022 MeV, which came from the original photon.*

The rest of the energy of the original photon is divided between the two newly produced electrons. For example, let us start with a photon of energy 5.022 MeV, or more generally E . The arithmetic would break down as follows:

	<i>This Example</i>	<i>Generally</i>
Total input energy:	5.022 MeV	E
Needed to make electron and positron:	1.022 MeV	1.022 MeV
Left over after pair is made:	4.000 MeV	$(E - 1.022) \text{ MeV}$

This leftover energy is divided, generally unequally, between the electron and positron as kinetic energy.

Pair production requires the presence of a nucleus to serve as a sort of catalyst. Although less frequently, an electron also can serve this purpose. In this interaction, the electron struck by the photon also acquires considerable speed, since it is driven forward by the photon. As a result, *three* rather than *two* high speed electrons emerge from the interaction (the electron pair plus the original electron). The process is therefore referred to as "triplet production." Note that triplet production requires that the photon energy be at least 2.004 MeV, twice as great as that required for pair production.

SUBSEQUENT EVENTS—ANNIHILATION. Both the positron and ordinary electron are fast moving charged particles. Therefore, they can ionize and excite atoms of the absorber and do so until they are slowed down to low energies. In this respect, they both behave similarly to photoelectrons and recoil electrons and give up their kinetic energy to the medium. Hence, in general $(E - 1.002) \text{ MeV}$ is absorbed. It should be noted that the initial energy of each electron and positron is usually rather high, and secondary ionization may take place several millimeters from the original collision in tissue.

* $E = mc^2$. Here "m" is the mass destroyed, "c" the velocity of light, and "E" the energy produced, all in consistent units. The identity applies as well for energy transformed into matter.

After slowing down, the electron acts like any other and simply increases the number of free electrons in the medium by one. The positron, however, is unstable at low speeds. It is an unfortunate stranger to our ordinary universe and cannot exist very long near an ordinary electron without their mutual extermination! This ill-fated union is referred to as "annihilation." The process is indicated in Figure 13 (Bottom). The positron, after slowing down sufficiently, interacts with the nearest electron. (In general, this will not be the particular electron called into existence at the time of the positron's birth because the pair initially moved in more or less opposite directions.) The two particles interact and are destroyed, and in their place appear two photons, traveling in opposite directions, each having 0.511 MeV of energy. The 1.022 MeV total energy is exactly equivalent to the total mass of the positron and electron annihilated. In this reaction, the converse occurs of what happened during pair production: matter is destroyed to produce energy. There is some element of poetic justice here. The positron owes its existence initially to 0.511 MeV of photon energy; it ceases to exist by repayment of the debt, restoring the 0.511 MeV in full as a new annihilation photon.

OVER-ALL SIGNIFICANCE. What is the result of these events? The absorber in which the annihilation and pair production occurred receives $(E - 1.022)$ MeV energy (absorption), primarily from the original pair. In addition, the net result of the annihilation is release of two oppositely directed 0.511 MeV photons. These are quite penetrating, and generally constitute deflection attenuation. Some annihilation photons irradiate extensive areas beyond the original field. In general, this can present a safety problem and in therapy increases patient volume dose. However, these effects are not serious in clinical practice because pair production is not a major mechanism in soft tissue irradiation.

Table 4 shows a comparison of the relative significance of Compton effect and pair and triplet production in water at various photon energies. Note that triplet production is far less important than pair

TABLE 4. FRACTION OF THE TOTAL ATTENUATION IN WATER DUE TO COMPTON EFFECT, PAIR AND TRIPLET PRODUCTION

Photon MeV	Percent of Total μ/ρ Due to:		
	Compton	Pair Production	Triplet Production
5	92	8	0.1
10	78	21.1	0.9
15	66.4	31.8	1.8
20	44.5	51.7	3.8

production, and both are normally far less important clinically than Compton absorption. Pair production increases with atomic number and rises slowly with photon energy above 1.022 MeV.

Photonuclear Disintegration

This type of reaction is far less important clinically than all four others mentioned and is included primarily for completeness. The effect involves a resonant type of interaction between the photon and the nucleus. It is roughly analogous to the photoelectric effect, which was a resonant interaction between the photon and inner orbit electrons. As in the photoelectric effect, the photon is completely lost and its energy given to the particle involved. In this case, however, the particle is either a photoneutron or a photoproton. (Examples are oxygen peaking at 24.2 MeV and carbon at 22.9 MeV, respectively.) The ejected heavy particles can interact with atoms they strike and produce ionization.

The initial atom is changed into a new isotope or element by this type of interaction, depending on which particle is ejected. More generally, photonuclear reactions may involve other types of products besides protons and neutrons, but these are beyond the scope of this discussion.

ATTENUATION COEFFICIENTS

Our discussion has been qualitative so far, dealing primarily with concepts and definitions. It is necessary to discuss attenuation *quantitatively* as well, in order to apply these principles to practical x-ray applications. In particular, we must examine how attenuation coefficients vary with Z and photon energy.^{3, 4}

We now consider some basic aspects of attenuation coefficients, their variation with Z and photon energy, and their application to design of filters and diagnostic contrast media.

Some Basic Aspects

As previously mentioned (p. 123) the mass attenuation coefficient (μ/ρ) more uniquely characterizes the attenuation by a given material than the linear coefficient (μ) because it is independent of absorber density. This is illustrated in Table 5, which compares the coefficients of ice, water, and steam. Note that μ/ρ is the same for all three forms of water. However, the μ values vary greatly with the form, reflecting changes in absorber density. For this reason, pub-

TABLE 5. LINEAR VS. MASS ATTENUATION COEFFICIENTS OF WATER FOR ITS THREE STATES (40 keV PHOTONS)

Form	Density	$\mu/\rho - \text{cm}^2/\text{g}$	$\mu - /\text{cm}$
Liquid	1.000 g/ml	0.264 cm^2/g	0.264/cm
Ice	0.917 g/ml	0.264 cm^2/g	0.242/cm
Vapor at 100° C	0.000598 g/ml	0.264 cm^2/g	0.000158/cm

lished coefficients are always μ/ρ values; where μ is desired it can then be calculated by multiplying μ/ρ by the absorber density ρ .

Two or more attenuation mechanisms usually act together to produce the total observed. Since there are four important mechanisms, the observed μ/ρ value results from the summation of photoelectric, the two types of scatter, and pair production effects. (Photonuclear disintegration and other mechanisms are ignored in this discussion.) Study shows that we can obtain coefficients corresponding to μ/ρ which describe the attenuation contributed by the separate mechanisms. These vary individually with absorber atomic number and photon energy, as do the effects they represent. Table 6 indicates the terminology used to indicate the total and partial linear and mass attenuation coefficients.

TABLE 6. SYMBOLS FOR LINEAR AND MASS ATTENUATION COEFFICIENTS

Attenuation Mechanism	Linear Coefficient	Mass Coefficient
1. Photoelectric effect	τ	τ/ρ
2. Scatter	σ	σ/ρ
a. Simple	σ_c	σ_c/ρ
b. Compton	σ_i	σ_i/ρ
3. Pair Production	κ	κ/ρ
Total	μ	μ/ρ

We have previously qualitatively described the variation of the various mechanisms with the absorber atomic number and photon energy. Table 7 indicates these relationships somewhat more precisely. Both types of scatter are essentially *independent* of Z , while photoelectric effect is roughly proportional to Z^3 and pair production to Z^2 . Thus Z is of importance primarily at very low energies (diagnostic range) and very high energies (betatron therapy). Compton scatter also varies very slowly with photon energy; on the other hand, photoelectric attenuation decreases very rapidly with increasing photon energy. Simple scatter attenuation decreases somewhat less rapidly

TABLE 7. HOW THE COMPONENT ATTENUATION COEFFICIENTS VARY WITH ATOMIC NUMBER AND PHOTON ENERGY E

1. **Photoelectric Effect**— $\tau/\rho \propto (Z/E)^3$
2. **Simple Scatter**
 σ_c/ρ rises *slowly* with Z.
 σ_c/ρ falls with E, as about $E^{1.7}$, for H₂O.
3. **Compton Scatter**
 σ_i/ρ is substantially independent of Z.
 σ_i/ρ falls very gradually with E.
4. **Pair Production**
 κ/ρ is roughly proportional to Z^2 .
 κ/ρ is zero below 1.022 MeV. It rises slowly with E above 1.022 MeV.

than photoelectric absorption. Pair production is zero below the threshold value of 1.022 MeV and rises slowly beyond that.

Curves of Mass Attenuation Coefficient versus Photon Energy

GRAPHICAL PROCEDURE. μ/ρ could be plotted against photon energy on three kinds of graph paper: ordinary linear, semilogarithmic, and log-log paper (Fig. 14). One is usually interested in a wide range of photon energy, from 10 keV through at least 1,000 keV. This results in μ/ρ values ranging from 5.31 cm²/g to .0706 cm²/g for water. For metals, μ/ρ varies even more.

To include all the desired data in convenient form on a single graph therefore requires logarithmic scale for both μ/ρ and E. Log-log graph paper fills this requirement. On any logarithmic graph scale, a 10:1 range is called a "cycle"—i.e., 10–100, 100–1,000 keV, etc. Thus, to cover from 10 keV to 10 MeV requires three cycles (10–100 keV, 100–1,000 keV or 1 MeV, and 1 MeV–10 MeV). To cover .001 to 10 cm²/g for μ/ρ requires four cycles. Another advantage of the log-log type of graph is that it results in a relatively straight line for the photoelectric attenuation (low keV) part of the μ/ρ vs E curve. This

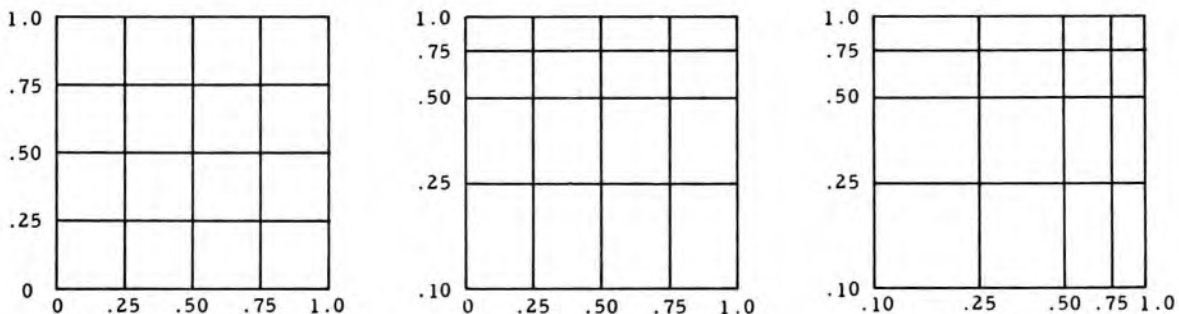


Fig. 14. Three types of graph paper scales. Left. Linear. Center. Semi-log. Right. Log-log.

is because the photoelectric coefficient versus photon energy relationship involves a power function (Table 7), and such a function yields a straight line when plotted on log-log graph paper.

μ/ρ VERSUS PHOTON ENERGY FOR WATER. Water is an extremely important absorber in clinical radiology, primarily because it constitutes about 70 percent of tissue weight in man. Biologic soft tissues generally attenuate x-rays very similarly to water, whose attenuation characteristics are hence of great interest.

Figure 15 presents a curve of μ/ρ versus photon energy for water. Calculations have been made and published for the component attenuation coefficients; these are also indicated by dashed lines. Pair plus triplet production coefficients (κ/ρ) are too small in magnitude to

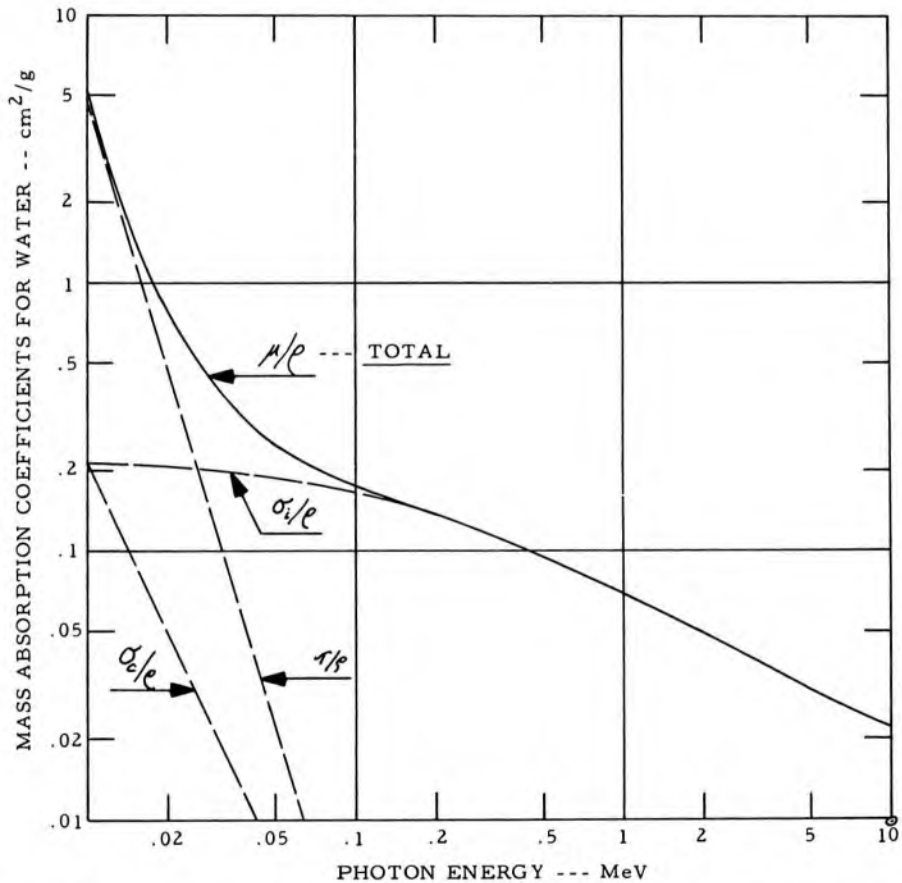


Fig. 15. Total mass attenuation coefficient (μ/ρ) vs. total photon energy, for water. Also shown as dashed curves are the component coefficient curves for simple scatter (σ_s/ρ) photoelectric absorption (T/ρ) and Compton scatter (σ_i/ρ). Note Compton scatter remains relatively constant with photon energy and is the dominant photon attenuation mechanism from 28 keV up to 25 MeV, virtually the entire range of clinical interest. Also shown is a point circled for the pair and triplet production coefficient (κ/ρ) of 10 MeV photons.

show on the graph (.0048 cm²/g and less). Note that at energies below 28 keV the dominant attenuation mechanism in water is the photoelectric process. At 10 keV there are equal amounts of Compton and simple scatter. Compton scatter is substantially constant over a wide range of energies. For example, it is reduced only 50 percent when the photon energy rises from 10 to 400 keV.

At 25 keV roughly half the x-ray attenuation of water is due to Compton scatter, the other half to the combined action of the photoelectric effect and simple scatter. However, the amounts of both photoelectric and simple scatter rapidly decline with increasing photon energy. As a result, substantially all attenuation in water is by Compton scatter from about 60 keV up to about 2 MeV because it is the only process operative to a significant extent. *Compton scatter is therefore the dominant attenuation mechanism in soft tissue for virtually the entire range of clinical x-ray work.* It is still dominant for water up to 25 MeV, when pair and triplet production catch up. Table 8 shows how much the Compton process contributes to the total x-ray attenuation by water from 25 keV through 25 MeV.

TABLE 8. FRACTION OF TOTAL X-RAY ATTENUATION BY WATER DUE TO COMPTON SCATTER

Photon Energy	25 keV	50 keV	100 keV	150 keV	200 keV	500 keV	1 MeV	2 MeV	5 MeV	10 MeV	25 MeV
Fraction of Total	50%	84%	96%	98%	99%	100%	100%	99.3%	92%	78%	50%

THE MASS ENERGY TRANSFER COEFFICIENT $(\mu/\rho)_{en}$.⁵ So far we have discussed (μ/ρ) which includes both deflection and absorption attenuation. μ/ρ is useful for computing narrow beam transmission through tissue and filters, and more generally for any application in which the fate of the absorber material is of secondary interest to its transmissive properties.

However, our main concern in many applications in radiology is with the energy deposited in an irradiated material. Absorption attenuation is then relevant, and the rest (deflection attenuation) essentially constitutes an undesirable loss of energy. These applications are usually the ones in which physical and chemical changes in the absorber are the objective of the irradiation. Examples are tissue, photographic film and fluorescent screens, and detectors in radiation measuring instruments. For such applications, a measure is desired of the *transfer of energy* to the medium, and the "energy transfer coefficient" is employed: $(\mu/\rho)_{en}$. Tables of these coefficients are given in N.B.S. Handbook 85.

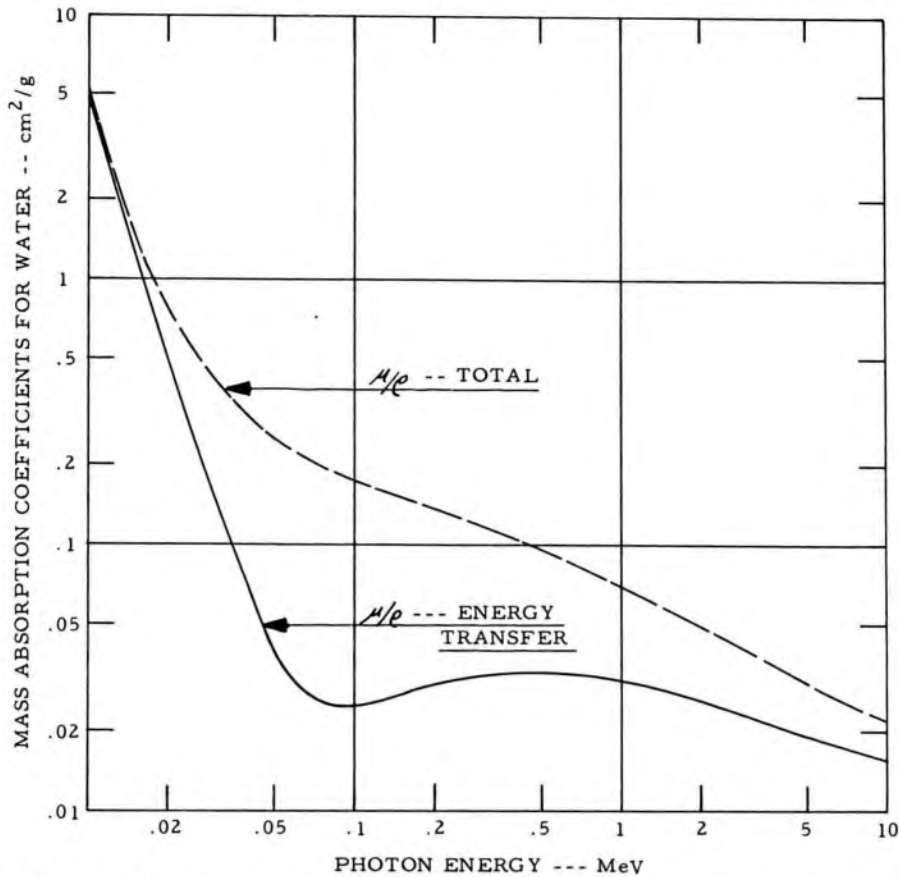


Fig. 16. Energy transfer mass attenuation coefficient for water vs. photon energy (solid curve). The *total* mass attenuation coefficient is shown for comparison (dashed curve). (Data taken from Berger.⁴)

Figure 16 shows (μ/ρ) and $(\mu/\rho)_{en}$ of water plotted against photon energy. Since $(\mu/\rho)_{en}$ is only a part of the total (μ/ρ) , its curve (solid line) lies below the total (μ/ρ) curve (dashed line). It is evident that at low photon energies water or soft tissue has considerable absorption attenuation. However, this is reduced very rapidly for energies above 20 keV, reaching a minimum at about 80 keV where deflection is maximum. This is unfortunately the general region (80–130 keV effective energy) of conventional deep therapy. Such beams can therefore be expected to deliver larger amounts of scattered energy outside the treatment field than higher energy beams, with consequently worse patient radiation illness (Chap. 7). The relatively high absorption attenuation above 700 keV results from more efficient energy transfer to the absorber by the Compton process at higher energies—i.e., the recoil electrons receive a higher fraction of the incident photon's energy (Table 3).

Table 9 compares the relative contributions of three interaction processes to $(\mu/\rho)_{en}$.

TABLE 9. FRACTIONS OF THE TOTAL ENERGY TRANSFER COEFFICIENTS OF WATER DUE TO VARIOUS MECHANISMS

Photon Energy	Fraction of the Total Mass Energy Transfer Coefficient in Water from:		
	Photoelectric Effect	Compton Scatter	Pair & Triplet Production
10 keV	.999 5	.000 5	NONE
15	.996	.003	↓
20	.986	.014	
30	.932	.068	
40	.795	.205	
50	.608	.392	
60	.432	.568	
80	.197	.803	
100	.090	.910	
150	.025	.975	
1.5 MeV	NEGLIGIBLE	.999	
2	↓	.993	.006 9
3		.953	.047 4
4		.933	.066 7
5		.900	.100
6		.862	.138
8		.796	.204
10		.730	.270
15		.601	.399
20		.530	.470
30		.384	.616
40		.303	.697

Notes:

1. Compton and photoelectric mass energy transfer coefficients are equal at about 57 keV.
2. Compton and pair plus triplet production energy transfers are equal at about 21 MeV.
3. The corresponding photon energy values for attenuation coefficients differ greatly from these figures: 25 keV and 25 MeV for photoelectric and pair and triplet production, respectively.

Applications

PURPOSE OF FILTERS. The beam emerging from an x-ray tube may contain photons whose energies range from the maximum to very low values. The low keV photons are attenuated very effectively by superficial layers of tissue because their coefficients are large. This is desirable for some superficial therapy but disadvantageous for both diagnosis and deep x-ray therapy work. In addition, there are many other applications in which it is desirable for x-ray beams to exhibit relatively constant attenuation as they traverse irradiated material. Filters are useful in these and deep therapy applications.

TABLE 10. NARROW BEAM TRANSMISSION BY WATER OF DIAGNOSTIC PHOTONS—PERCENT

Photon Energy (keV)	Thickness of Water—cm			
	5	10	15	20
15	0.0336	----	----	----
20	1.94	0.04	0.0008	0.000016
25	8.1	0.66	0.053	0.0043
30	15.8	2.5	0.4	0.06
40	26.8	7.2	1.9	0.5
50	33.0	10.9	3.6	1.2
60	36.1	13.0	4.7	1.7

Table 10 shows the narrow beam x-ray transmission by water or soft tissue of various monochromatic beams. It is evident that photons of 25 keV and lower energy are practically all removed by 20 cm of tissue before reaching an x-ray film and hence contribute negligibly to the diagnostic image. Twenty keV photons are correspondingly removed by 15 cm parts, and 15 keV photons, 5 cm parts. Hence, in most diagnostic x-ray work photons 25 keV and less contribute greatly to patient dosage but negligibly to the image for thicker parts; even thinner parts rather effectively remove photons up to near 25 keV. (However, in some particular studies like mammography low keV photons are employed to secure essential diagnostic information despite the need for large amounts of x-ray directed at the part. In mammography the presence of considerable fat in the breast also helps render it relatively transparent.)

In deep therapy one would like to remove photons of less than about 40 keV, to maximize the transmission of x-rays to deep tumors. Another benefit is the reduction in low energy photons in the beam; these are disproportionately injurious to skin (Chap. 7).

In short, the purpose of a filter is to selectively remove low energy photons which are of too low penetration to contribute efficiently to the diagnostic or therapeutic objective. Selection of filter material and thickness is a fairly sophisticated problem which will be considered below.

MATERIALS AVAILABLE. Filters are generally made of metals such as aluminum, copper, tin, lead, or gold. Figure 17 shows a graph of the total mass attenuation coefficients versus photon energy for the three most common of these materials: aluminum, copper, and tin. Note that at higher energies all the curves tend to merge. This is because Compton attenuation becomes dominant for all these metals at energies a few hundred keV and above. Recall that the attenuation per electron is substantially independent of atomic number in Comp-

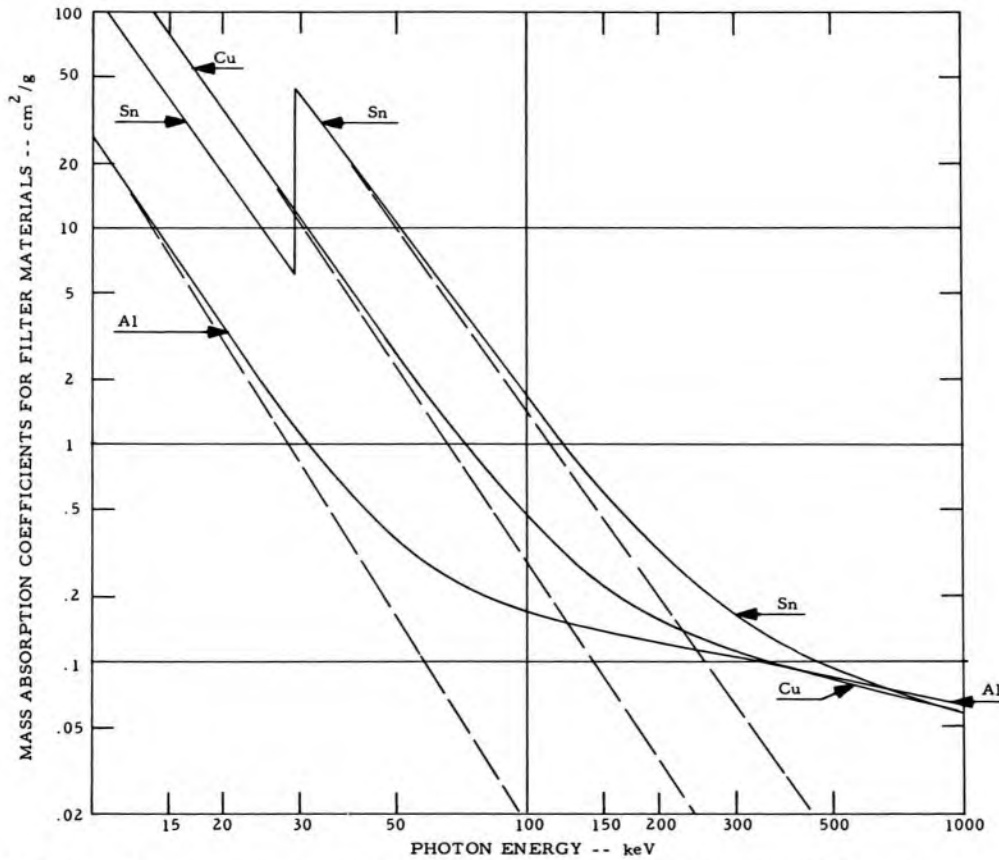


Fig. 17. Total mass attenuation coefficients (solid curves) for three useful filter materials: aluminum, copper, and tin. The dashed curves indicate contribution of the photoelectric process (T/ρ) to the attenuation coefficient, for each metal. The metals are generally most effective as filters for the ranges where the dashed lines approximate the solid curves.

ton interactions. For these metals, there is roughly the same number of electrons per gram of metal, so μ/ρ is also the same. At very high energies (not shown), pair production becomes significant, particularly for higher Z metals.

For purposes of comparison, the dashed curves show the contribution of photoelectric absorption (κ/ρ) from each metal. κ/ρ and σ_i/ρ are seen to be about equal for photon energies of 47, 124, and 245 keV for Al, Cu, and Sn, respectively. Note the very steep regions of the curves below these keV values.

FILTER PRINCIPLES.⁶ Ideally, a deep therapy filter should transmit *no* photons of energy below about 40 keV and *all* photons of energy above this value. To achieve this would require that μ/ρ be *infinite* below 40 keV and *zero* above 40 keV. This is impossible because no material known has this type of variation of μ/ρ with E . All we can ask is that there be a *maximum difference* in μ/ρ for the two keV values involved. This is achieved for the steepest slope of the μ/ρ curve

in Figure 17. For this part of the curve, a given keV interval (for instance, 30 to 100 keV) corresponds to the biggest difference in μ/ρ , and hence the greatest difference in x-ray transmission by the filter of the two photon energies. The steep slope corresponds in general to the low energy part of the metal's attenuation coefficient curve, where this slope approaches that of pure photoelectric effect.

Aluminum is used as a diagnostic x-ray filter. It has a ratio of μ/ρ between 20 and 50 keV of 3.48/0.36, or 9.7 times versus 13.9 times for copper. Aluminum is therefore, from a theoretical point of view, inferior to copper as an x-ray filter in this range. Unfortunately, a copper filter for diagnosis would have to be so thin as to be impractically fragile.

Copper and tin are normally used in deep therapy, since aluminum absorbs excessively by Compton scatter around 100 keV (approximately the effective photon energy for deep therapy x-ray beams). Even copper has considerable Compton effect at 100 keV and above (Fig. 17). For this reason, we use tin filters for deep therapy wherever possible. But, note that near 29 keV there is an abrupt drop in the attenuation coefficient of tin, making it inferior to copper as an absorber of very low energy photons.

PRACTICAL FILTER DESIGN. Before discussing specific filters, it may be useful to summarize the points developed above.

1. For diagnosis, aluminum provides the overall best filter material, usually of thicknesses of 2 to 3 mm, except in very low kV studies like mammography. It is preferred to copper because relatively thick and rugged filters may be used.
2. In orthovoltage therapy (140 to 320 pkV), maximum steepness of the μ/ρ versus photon energy curve is desired, so tin is generally best with copper second best as a filter material.
3. Tin, however, fails abruptly as a filter material below 29 keV. In clinical use it must be backed by some copper to take over at these low photon energies when tin becomes relatively transparent.
4. Tin and copper both produce their own soft characteristic radiation when irradiated (about 29 and 9 keV, respectively). They must both be backed by lower Z materials to remove this superficially absorbed radiation.

Consequently, multi-layer filters are used in orthovoltage therapy; they are often called, for obvious reasons, "composite" filters.

Figure 18 shows the basic location of filters in the x-ray beam and the arrangement of their components in three common designs. The first [Fig. 18(Left)] is the simple aluminum filter generally used in

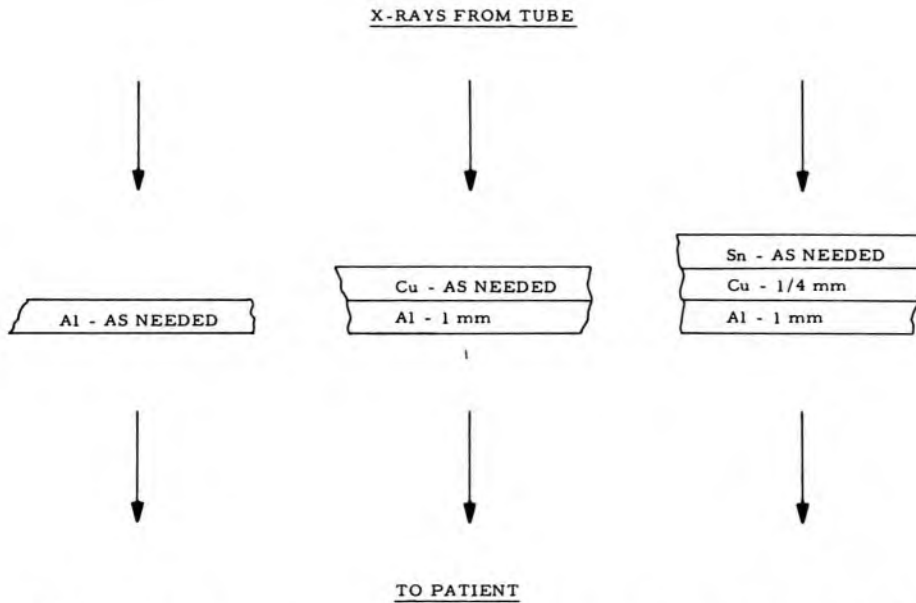


Fig. 18. Common arrangements of filters in x-ray machines. Left. Simple aluminum filter. Center. Copper filter. Right. Thoraeus filter. In copper and Thoraeus filters, lighter metals are used to attenuate x-rays in the discontinuity range of the adjacent heavier metal.

diagnosis. In superficial therapy, up to 4 mm of aluminum is also employed. For HVL values up to about 1.0 mm of copper, simple copper-aluminum combination filters are used [Fig. 18 (Center)]. One millimeter of aluminum is always provided to remove the 8 to 9 keV copper characteristic radiation.

These assemblies usually employ 0.25 to 1.0 mm of copper sections. Where greater filtration is required to achieve beams of higher HVL values, tin is usually added ahead of a 0.25 mm copper filter [Fig. 18 (Right)] and the assembly called a "Thoraeus" filter (in honor of the designer). For example, a Thoraeus-II filter yields the same beam HVL as a 2 mm copper, 1 mm aluminum filter. It consists of approximately 0.4 mm of tin followed by 0.25 mm of copper plus 1 mm of aluminum. If a greater filtration is needed, more tin may be used. Thus, a Thoraeus-III filter uses about 0.6, a Thoraeus-IV about 0.8 mm of tin, also with 0.25 mm copper plus 1 mm of aluminum. The beam HVL then is about the same as that of 3 and 4 mm copper plus 1 mm aluminum filters, respectively. In general, a Thoraeus filter produces about the same HVL as a corresponding copper plus aluminum filter but with greater beam intensity (for example, about 20 percent higher for a Thoraeus-II filter than that obtained using a 2 mm copper and 1 mm aluminum filter). The resultant saving in treatment time is of great importance in irradiating sick people for whom a protracted treatment can present quite an ordeal.

X-ray Attenuation Discontinuities

It is puzzling when one first observes the abrupt reduction in attenuation coefficient of tin at 29 keV (Fig. 17). This is usually referred to as the "attenuation discontinuity" of tin. There is an abrupt reduction of *six times* in the ability of tin to absorb x-rays, when the photon energy is reduced from 29.3 down to about 29.1 keV. This is very significant from the point of view of filter design, necessitating a back-up copper filter to avoid undue transmission of undesired soft radiation below 29 keV.

Such discontinuities occur with all materials in the photoelectric region. If one continued the graphs of Figure 17 below 10 keV, he would find similar discontinuities to that of tin at about 9 keV for copper and 1.55 keV for aluminum. In general, the discontinuity occurs at higher photon energies for higher atomic number materials. For example, the discontinuity of lead occurs at about 88 keV and of tungsten at about 69.5 keV. Note that these are just above the K-characteristic photon energies for all these materials; they constitute the K-orbit electron binding energies.

PRACTICAL SIGNIFICANCE. We have mentioned the significance of attenuation discontinuity in filters. In addition, attenuation discontinuities also contribute to selection of barium and iodine as preferred contrast media in x-ray diagnosis. The discontinuities of higher atomic number materials occur in an embarrassing location, as far as x-ray diagnostic work is concerned. For example, suppose one had some very well-bound lead compounds, chelated or otherwise processed so they were essentially impervious to chemical breakdown in the human body. One might think offhand such compounds would be an excellent contrast medium for x-ray diagnosis because of the high Z of lead. Lead, however, attenuates inefficiently below 88 keV because of its discontinuity. Since the region below 88 keV is crucial for effective x-ray diagnosis, lead is generally a poorer contrast agent, gram for gram of element, than barium or iodine. For example, its μ value for the same concentration by weight in a body cavity is only 0.7 times that of iodine between 33 and 88 keV.

CAUSE. The attenuation discontinuity results from the resonant character of the photoelectric effect. This mechanism involves primarily inner orbit electrons and is a maximum for photon energies just greater than the binding energy ϕ . As one reduces photon energy, all orbits absorb better, reaching a maximum near ϕ . For tin, at 29.2 keV, the K orbit absorbs most effectively. The photoelectrons then barely leave the atom since ϕ is just exceeded by the incoming photon energy. However, when the photon energy is *less* than 29.2 keV, the K-

electrons abruptly do not receive enough energy to escape the atom, so the K orbit no longer attenuates the beam. As a result, only the remaining orbits participate (L, M, etc.). These absorb relatively weakly, so μ/ρ is reduced to only one-sixth of the value above 29.2 keV. To summarize, K-orbit electrons interact photoelectrically only when their threshold energy is exceeded. Otherwise, they do not respond to the incoming photons, and the atom attenuates x-rays without K orbit participation. This results in the observed discontinuity.

Similar discontinuities also occur for the L orbits, but these are at much lower energies and unimportant for elements of Z values up to that of tin in clinical radiology.

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5

Biologic Effects of Ionizing Radiation

This and the next two chapters deal with the action of ionizing radiation: its effects, measurement, and therapeutic application.

X-rays and other ionizing radiations produce profound biologic effects that are essentially destructive. In the treatment of malignant disease, x-rays inevitably damage some healthy tissue irradiated in the process. In diagnosis and other applications, incidental hazards to patients, personnel, and “innocent bystanders” must be minimized by careful planning procedures. It is evident that an understanding of the production of radiation injury is of basic importance in applying ionizing radiation.

In this chapter we shall discuss how x-rays interact with tissues and what variables affect the result.

PRODUCTION OF RADIATION REACTION

Certain obvious considerations affect the response to *any* physical agent striking an organism. These include the intensity, duration, and extent of the area exposed. For example, in exposure to heat, ultraviolet, or radar beams, the intensity and duration directly affect the degree of *local* injury (skin and superficial tissues); in addition, if an excessive area of the body is involved, organism survival may be jeopardized. Similar considerations apply to x-rays as well.

However, x-rays also have unique properties contributing to other, more special types of injury. First, their great penetration delivers injurious radiation deep in the body: to reproductive cells, blood-forming organs, intestinal mucosa, etc. In addition, x-rays are very insidious in action because they produce no sensory response at usual intensities and there is delayed action or “latency.” This latter term refers to the delay in appearance of ultimate radiation effects, anywhere from *days* to *years* following irradiation.

Sequence of Events

As shown in Chapter 4, the attenuation of x-rays involves primary interaction with electrons and nuclei of atoms, most generally with the release of fast electrons. These electrons, in turn, produce secondary ionization of other atoms and thereby chemical changes in cells. Ordinary protective mechanisms of the body cannot be effective against injury produced so specifically within cells. While the question is not as yet completely settled, the major injury is apparently to the reproductive mechanism of cells. There can be aberration of the reproductive integrity (mutations) and loss of the ability of the cell to reproduce altogether (sterilization). When the dosage delivered is sufficient, the cell may be destroyed outright, but this is normally far less common than chromosomal damage.

Table 1 shows the basic sequence of events in production of biologic injury.^{1, 2} The physicochemical stage refers to the production of ionization and excitation and the immediate aftermath. Note this takes place in much less than one second! The biologic stage refers to the organism's reaction to the radiologic lesion. So far, unambiguous experimental results are available primarily for relatively intense irradiation, and Table 1(B) is limited to this.

THE RADIOLOGIC LESION. The action of primary electrons (or more generally fast-moving charged particles) is to produce ionization and excitation of atoms with which they interact. The ions and excited atoms then initiate chemical changes. In the living cell, the action can be direct or indirect.

Direct action occurs when essential molecules are altered directly by the ionization process: i.e., enzymes, DNA molecules, etc. This is the major mechanism of beams of *heavy* particles, like neutrons, fast protons, and alpha rays, but *not* x-rays. In clinical situations, electrons leave a trail of ions of substantial average separation, rather uniformly distributed in the cellular volume. Consequently, the ions rarely react with vulnerable cellular sites intensively enough to effectively denature them (see p. 158).

Indirect action first involves production of free radicals in water (H^{\bullet} and OH^{\bullet} directly, and other intermediary radicals). These survive long enough to migrate by diffusion and to initiate chemical action at relatively remote vulnerable locations (up to 100 microns).

The net effect of all chemical changes is twofold: essential chemical constituents may be altered so their function is lost, and new toxic constituents are produced.

SUBSEQUENT EVENTS. The initiating ionization and excitation events occur in less than 10^{-12} second (a *millionth of a millionth*

TABLE 1. STAGES IN PRODUCTION OF RADIATION INJURY
Physiochemical Stage¹

Step	Time Involved
1. Passage of particle (primary electron and delta rays* ionize atoms)	10 ⁻¹⁶ second
2. Initial chemical action	10 ⁻¹² second
a. directly on cellular molecules	
b. formation of free radicals	
3. Intermediate effects in aqueous parts of cell	10 ⁻⁷ second
a. diffusion of H [•] and OH [•] radicals from track (x-rays)	
b. intra-track recombination to produce H ₂ O ₂ with high LET rays (protons, alpha rays, neutrons)	
4. Indirect action on cellular molecules in H ₂ O	10 ⁻³ second
5. Oxygen effect	2 × 10 ⁻² second
6. The entire radiologic lesion	less than one second!

Biologic Stage (for higher doses)^{† 2}

Step	Time Involved Within:
1. Cellular damage may become apparent	1 hr
2. For whole body irradiation:	
a. bone marrow depression apparent	4 days
b. intestinal mucosal injury apparent	3 weeks
3. Visible erythema (local skin response)	8 days
4. Latent injury	few up to 30 years

Several causes contribute, including

- a. direct injury to irradiated cells, especially chromosomal.
- b. toxic effect from cellular death—local and systemic reaction.
- c. functional loss from cellular *death* (capillary loss, etc.)
- d. functional loss from cellular *sterilization* (intestinal mucosa, etc.)

* Delta rays are fast electrons released from atoms struck by a primary charged particle like an electron, proton, alpha ray, etc. After release these delta rays ionize other atoms on their own. For example, a particular 50 keV primary electron might use up 10 keV in actually ionizing atoms; its other 40 keV is delivered to the released electrons (delta rays), which use up this kinetic energy ionizing other atoms.

† The biologic response varies greatly with the dosage, volume irradiated, type of radiation, and many other physical, chemical, and biologic factors.

of a second). Even the slower resulting chemical changes take place in less than 1/50 second. However, the response of the individual cell, the tissue surrounding it, and the organism as a whole is much slower, and gross injury requires from several minutes up to years to become evident. The initial delay arises from the fact that cells are mainly sterilized, so the consequences of deterioration and loss of function of cells are not grossly evident during their lifetime. During the latent period, biochemical changes in the cell nucleus develop the hidden into manifest cellular and tissue injury. (This is crudely analogous to the development of a latent photographic image.)

VARIABLES AFFECTING RADIATION INJURY

While ionization is the basic cause of radiation injury, many additional factors also affect the final biologic result. These can generally be classified as physical, biologic, and chemical in nature. They vitally affect radiation therapy and protection, and a perusal of references on the subject is rewarding. While a full discussion of these factors is beyond the scope of this book, we shall review them briefly below.

Physical Variables

Three physical variables crucially influence the biologic injury from irradiation:

1. The extent and distribution of ionization within tissue.
2. The time sequence of treatment—both overall duration and spacing of treatments.
3. The type of radiation employed: if x-rays, the HVL; if particles, their type and energy.

All of these factors can be controlled to a considerable extent in treatment planning, so they are of great practical interest.

EXTENT AND DISTRIBUTION OF IONIZATION IN TISSUE. The local concentration of ions in tissue can be illustrated by two small cubic volumes of tissue 1 mm on each side [Fig. 1 (Top)]. The right has twice as many ions as the left for the same volume, so its cells on the average receive about twice as great a radiologic lesion. Thus a unit of local dose, or initial *physical* trauma, might be ion pairs per mm³ of tissue. Actually, excitation also takes place, so the *total* delivered energy per given amount of tissue is more directly applicable. The rad unit has been defined as an absorbed dose of 100 ergs per gram of irradiated material and is generally employed as the unit of local tissue energy deposition. As we shall see below, the ionization produced in a gram of tissue by

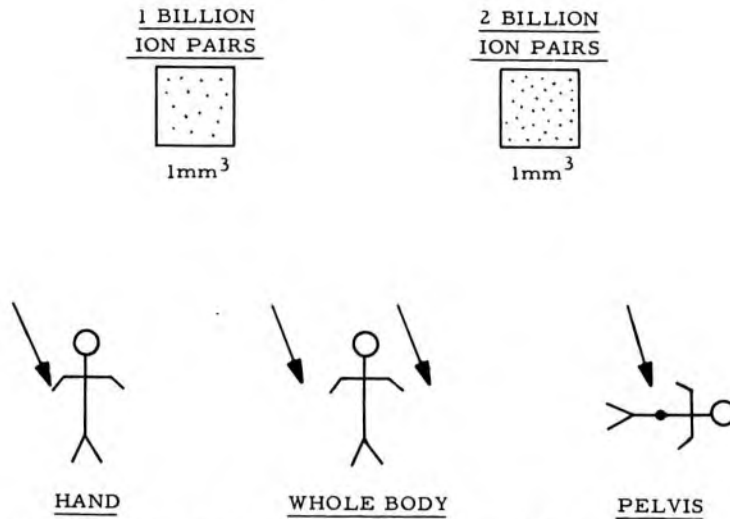


Fig. 1. Distribution of ions in tissue. Top. Roughly $\frac{1}{2}$ and one rads in soft tissue irradiated with cobalt-60 rays. Bottom. Local vs. extensive irradiation: to hand, whole body, and pelvis.

a given amount of energy varies with Z and the primary electron energy. It is nevertheless conceptually helpful, for most practical purposes, to visualize a rad as roughly two billion ion pairs in a 1 mm volume of soft tissue.

So far we have spoken of ion concentration at a single location, or rads. A second companion aspect refers to the rad doses in the various parts of the irradiated organism. Figure 1 (Bottom) shows three grossly different illustrative situations. In (1) a very small part of the patient is irradiated, such as in treatment of a basal cell carcinoma. (2) shows irradiation of the entire person ("whole-body irradiation"), the opposite extreme. Generally, more limited but still fairly large tissue volumes are irradiated in treating deep-seated lesions, such as the pelvic location indicated in (3). Obviously, for a given rad dosage, the fewest cells are injured in (1), the most in (2), and an intermediate number in (3).

From the point of view of organism survival, the total energy delivered is most critical. This helps determine the total cellular injury, and consequently the loss of tissue function and release of toxic products of tissue destruction. In both accidental irradiation and radiation therapy, delivery of large amounts of energy can produce radiation illness. The total energy deposited in the organism is referred to as "volume dose" and has a unit of *gram-rads*. A gram-rad is the energy delivered to one gram of tissue when it receives an absorbed dose (i.e., the accepted term for rad dosage) of 1 rad. By definition of the rad, this is numerically 100 ergs. A larger unit is often convenient, which is a million times greater, called the "megagramrad." (The calculation of volume dose is discussed in Chapter 7.)

In radiation therapy one usually desires to achieve the best "distribution" of dosage in and around the tumor. Most therapists agree that generally a relatively constant rad dosage should be given to all the tumor cells to assure adequate tumor control without excessive irradiation of contiguous healthy tissues.³ In addition, the lowest practical dose should be delivered to other more remote healthy tissues. Treatment planning therefore can involve much effort to assure optimum rad dosage distribution in and around the tumor.

Thus far we have spoken of local concentration of energy in tissue (rads) and overall distribution of rads in the body. These describe the situation grossly (as low as 1 mm³ volume). However, on a microscopic scale more subtle effects are involved. These result in a given rad dose of one kind of radiation producing a more destructive effect than the *same rad dosage* of another kind! For example, to effectively treat a certain tumor in 40 days requires a uniform dose of 6,000 rads, using a 250 kV beam (2.5 mm Cu HVL). If one uses cobalt-60 radiation instead, the actual rad dose must be raised to about 7,000 rads, even when all other variables are the same! Further examples will be given below and this entire subject discussed more fully in Chapter 7.

TIMING OF IRRADIATION. Treatment scheduling can be varied in several ways. The entire treatment can be given in a single relatively short period; this is referred to as a *massive* or *single-dose* technique. Alternatively, radiation can be delivered more slowly, in a period extending over several days or weeks. One then says the treatment is *protracted*. This is the usual situation in radiation therapy. Protracted irradiation is usually given in small fractions spread out over several weeks at daily or similar intervals. This is the procedure employed most commonly in x-ray therapy and is called *fractionated* technique.

One might ask, "Why not deliver the dosage all at once and get it over with?" This procedure certainly has practical advantages in reducing the cost of treatment and hospitalization, and it is sometimes used where preferred protraction is impractical. However, there is always greater sacrifice of healthy structures with massive irradiation. With this technique the tolerable dose is more critical and if exceeded can result in what is essentially radiation cautery. Massive treatment of small skin lesions is usually followed by a poorer cosmetic but still successful result. However, in deep tumors one must almost always extend the treatment time. This increases the required total dosage, but repair of healthy tissues is greatly facilitated.

The longer the overall treatment time, the greater the total rad dosage required to produce almost *any* biologic effect. Figure 2 shows the experimentally observed relationship between the total dose re-

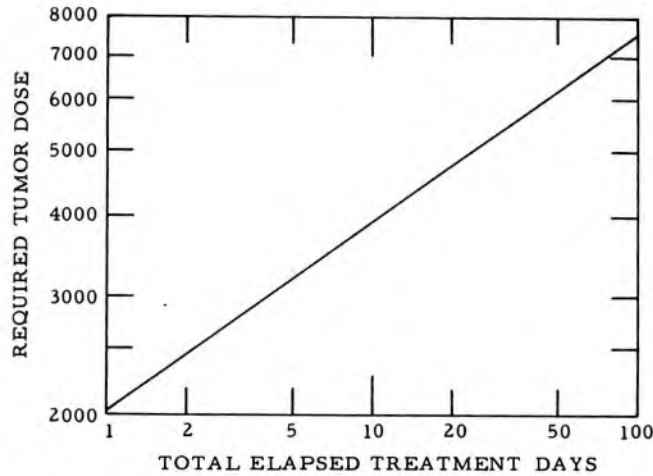


Fig. 2. Dosage needed to control a tumor vs. duration of treatment. This is based on Strandquist's data⁴ for skin cancer. Different curves have been reported by Paterson and Friedman, showing some curvature of the curves and different values. This serves to illustrate the complexity of cancer therapy.

quired to control a given tumor type and the total elapsed treatment days. (Strandquist's data are shown. Those of other investigators are similar, but differ significantly in both slope and dose magnitude, due to clinical variables.)⁴ As indicated above, the required total rad dose rises with elapsed time. Note both scales are logarithmic, to make the curve relatively straight.

For many years people have explained this effect by saying tissues "recover" between treatments. To illustrate, it takes 2,100 rads to control a tumor with a single dose, 2,500 rads (1,250 each on two successive days) to do so with two treatments. Were there *no* recovery, one would need only $1,250 + 850 = 2,100$ rads total. Hence, the day's rest has permitted tumor recovery to the extent that 400 rads *extra* is needed the second day to complete the job.

More recent work on cell cultures rather than intact animals has provided a clue to one of the recovery mechanisms.⁵ Radiobiologists are able to experimentally irradiate many kinds of cell cultures to various dose levels and measure the fraction of cells surviving. ("Surviving" here means retaining reproductive capacity.) In Figure 3, A curve is typical for HeLa cells, with the fraction surviving following irradiation plotted on semilog graph paper against dosage in rads. Note that a straight line results, except for an initial curved portion at low dosage levels. This implies the cell population can tolerate radiation more at low than at high dosage. For example, the first 200 rads drop the population by a factor of 0.3; the next 200 rads (total of 400 rads) reduce it from 0.3 to 0.07 of the original number. It appears a certain amount of radiation is needed to "prime" the

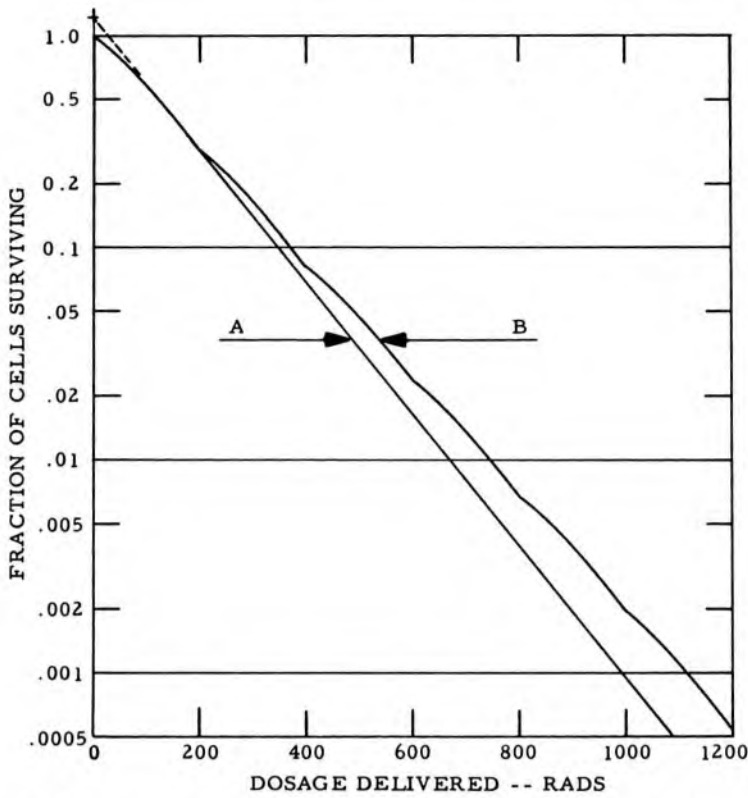


Fig. 3. Survival curves, computed for HeLa Cells ($D_{37} = 140$ rads, $n = 1.5$) in cell cultures.⁵ Curve A: single exposure. Note curve becomes almost perfectly straight after 150 rads exposure, but an initial dosage is required to “prime” cells. For continuous irradiation at high rate, cells cannot recover to re-establish their initial relative insensitivity, so survival falls more rapidly than initially. Curve B: fractionated exposures, 200 rads each. Note that after 6 fractionated sessions (1,200 rads) about four times as many cells survive as for 1,200 rads delivered in a single session. Note: This is for cellular cultures only and assumes overall treatment times are short compared with cell life span, etc.

cells and make them more vulnerable to radiation; after this is done they become maximally vulnerable, and a given increase in dose sterilizes a larger and constant fraction of those present.

It is believed interrupting the treatments usually results in recovery of cells to the point at which they must again be primed. The over-all effect of interruption is therefore to render cells less sensitive to radiation. Curve B in Figure 3 indicates the effect of daily fractionated doses of 200r on HeLa cells.

It is evident from the above why the total required tumor dose might be expected to increase with overall treatment time: essentially rads are wasted repetitively in “priming” cells. However, this does not answer other vital and interesting questions, such as why the extension of treatment time often permits more complete healing of healthy tissue without equal healing of the tumor? Is there an optimum treatment schedule for certain types and sites of tumors

which differs from that of others? These and many other questions are receiving active attention from radiobiologists; the forthcoming answers may make basic contributions to more effective radiation therapy.

THE TYPE OF RADIATION USED. Heavy particles are stopped rather quickly compared to electrons of the same energy, so their ions are more packed together, along relatively short paths (Fig. 4). This greatly influences the biologic response from a given amount of radiation energy in tissue. For example, a given rad dose of α -rays, neutrons, and proton beams generally produces much greater biologic effects than the same rad dose from either electron or x-ray beams.

For photon beams, the response of tissues to x- and gamma rays depends upon the beam quality; this dependence is apparently greater for skin than for any other type of tissue. Table 2 shows the variation of "threshold erythema dose" with beam quality.^{6, 7} Threshold erythema dose (TED) is defined as the skin dose required to produce a threshold erythema (faint reddening) in a small localized area of the forearm in 80 percent of a large group of people within three weeks. The threshold erythema dose is now considered to be much less

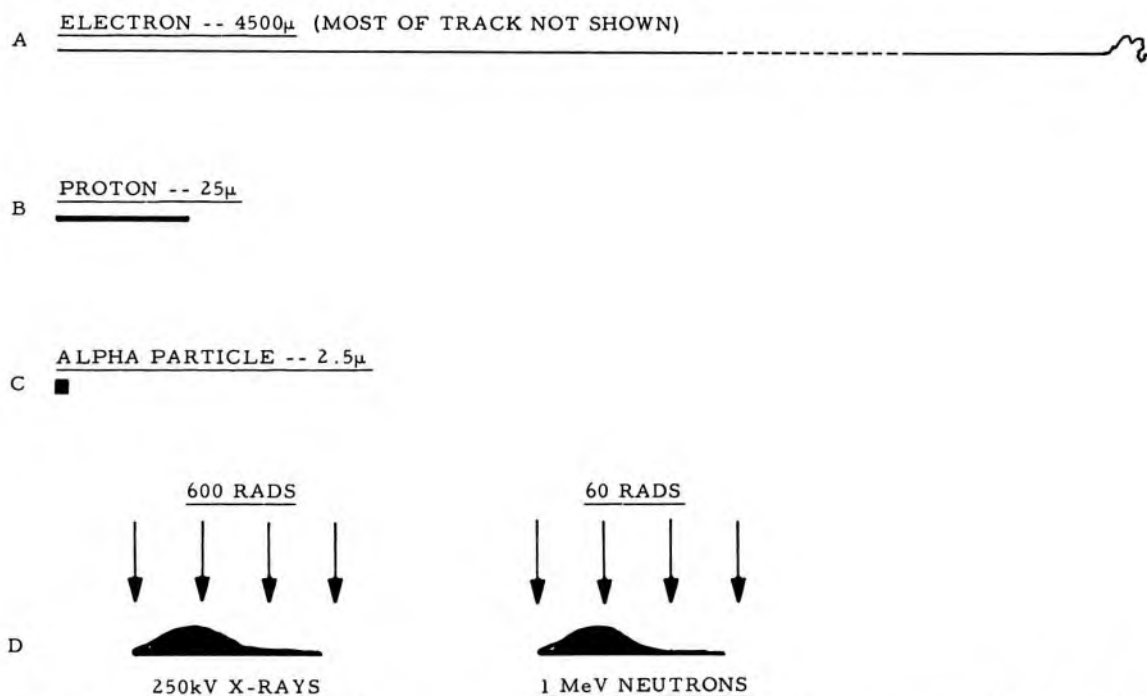


Fig. 4. The type of radiation used affects the biologic response. A, B and C compare the lengths of travel of three types of particles, all with the same energy (1 MeV). A. *Electron*: travels about 4,500 microns, or 4.5 mm. B. *Proton*: travels only 25 microns, or .025 mm. C. *Alpha particle*: travels only 2.5 microns, or .0025 mm. D. Here are shown mice exposed to lethal irradiation from orthovoltage x-rays and fast neutrons (1 MeV). Note the required rad dose is only one-tenth as great for the neutrons!

TABLE 2. THRESHOLD ERYTHEMA DOSE (TED) VALUES FOR X- AND GAMMA RAYS^{6,7}

pkV	mm HVL	TED Roentgens (Quimby)	Skin \bar{F} (Johns)	Estimated TED-rads (Johns)
100	1.0 Al	270	1.21	330
140	0.4 Cu	525	1.18	620
200	0.9 Cu	680	1.06	720
700	7.0 Cu	800	0.97	780
1,000	3.8 Pb	1,000	0.96	960
Radium	11.0 Pb	1,050	0.957	1,000

unique a biologic unit than formerly believed. However, it is generally agreed that a large TED is a good indication that minimal skin reaction will occur in clinical radiotherapy. It should be emphasized that the TED differs from the *erythema dose*, which is a loosely defined term meaning the typical radiation dose required to produce an erythema and with normal fractionation in a clinical procedure.

Table 2 indicates there is a tremendous dependence of skin response on photon energy. It is evident that a factor of 3 separates soft from hard radiation in terms of skin tolerance! This is one reason why appropriate filters must be employed in x-ray machines to remove the softer low energy radiation before it reaches the patient. If this is not done, adverse skin reactions can limit the attainable dosage at depth in radiotherapy and the permissible number of x-ray examinations in diagnosis.

It appears to be generally accepted that higher energy photons are less effective in producing biologic changes than lower energy photons. For example, it takes *more rads* of cobalt-60 supervoltage than orthovoltage to produce a given biologic effect, with all other variables the same.

Why should higher energy photons exhibit reduced biologic effects? The reason probably lies in the microscopic distribution of ions. These ions are produced along the paths of electrons released by photon interactions. In general, the more energetic the photons, the faster these released electrons. Faster electrons leave trails less densely populated with ions than slower ones. Figure 5 illustrates this, comparing reasonably typical primary electron tracks of an orthovoltage (A) and a cobalt-60 (B) beam. It is evident that a 0.6 micron diameter cell organelle is acted upon by far more ions from primary electrons of an orthovoltage beam (A) than from a cobalt-60 beam (B). Of course, for a given rad dose, the total number of ions produced in a mm³ is the same in both cases. However, more cells are *heavily ionized*

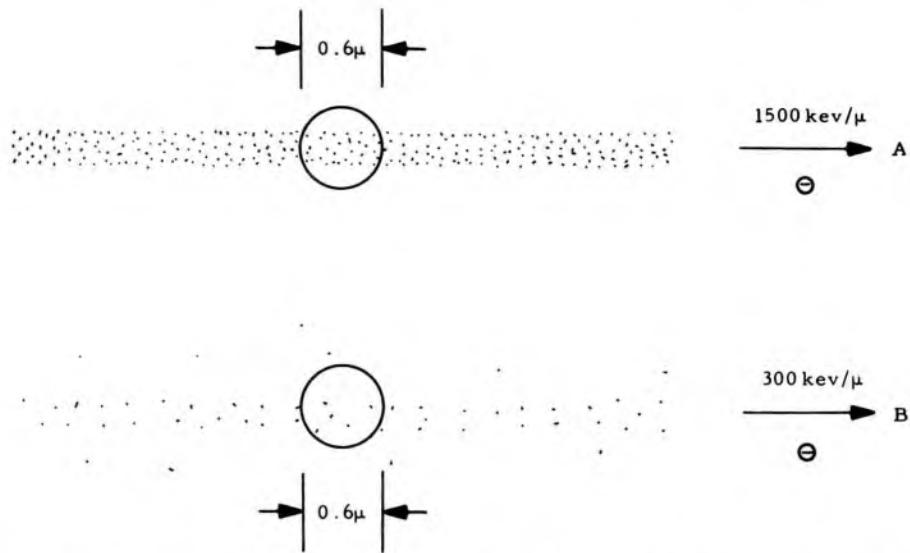


Fig. 5. Linear Energy Transfer (LET).⁷ LET is a measure of the ion density in the track of a charged particle. In x-irradiation, this would be the recoil or photoelectron released initially. The LET is always greater for slower moving charged particles because they remain near atoms longer, increasing the probability of ionization. A. 200 kV x-rays. The ions are shown as produced along a typical electron track in a cell. A hypothetical object 0.6 microns in diameter is seen to contain many ions from this one particle (1,500 keV/μ). B. Cobalt-60 rays. Corresponding typical lower LET ionization track (300 keV/μ).

Considerations of this sort are involved in the 0.85 RBE figure for cobalt-60 radiation (chap. 7).

by the orthovoltage beam; its electron tracks are consequently fewer, but with the ions packed closer together.

There is evidence many cells recover when injured at only a small number of sites but may be irreparably damaged when the number becomes excessive. Therefore, the likelihood of permanent cell injury increases when ions are more densely packed. The term "linear energy transfer" (LET) is often used in discussing this distribution of ions in the trail of the primary electrons. The linear energy transfer or LET is defined as the density with which energy is deposited along the track of the ionizing particle, usually in keV per micron.

While the LET concept helps explain the variation of biologic response with photon energy, it must be stressed that the picture is much more complicated than presented here, and the above discussion must be considered a very brief introduction to a complex subject.

Biologic Variables

Since a detailed consideration of the purely biologic aspects of this question is beyond the scope of this book, these general comments are in order. The biologic effect of a given irradiation procedure

varies greatly with the type as well as the particular animal chosen. Even among animals of the same breed, there is a wide range of dosage required to produce a given biologic effect, such as whole body damage, cataract, and tumor response. In any particular animal, the general state of health is also a factor.

With regard to treatment of a specific tumor in a specific individual, the response will depend to a very great extent on previous trauma and infection. Previous irradiation, surgery, or burning all adversely affect the healing properties of healthy tissue as well as the responsiveness of the tumor. Associated with this sort of situation is the circulation status, particularly in the region irradiated. Not only the state of the vessels themselves but also the general hematologic state is involved. Any anemia gravely affects the results of treatment by reducing oxygen tension in irradiated tissues. In addition, other related factors affect the biologic response to radiation.

Chemical Variables

The chemical environment of a cell profoundly affects its response to radiation. Agents are available which enhance or reduce this response; they are called *potentiators* and *protectors*, respectively. In radiation therapy one would like both to protect healthy tissues from radiation injury and to potentiate the response of tumor tissue.

Significant differences exist naturally among tumor and other tissues in both radiosensitivity and recovery rate after irradiation. Bergonie and Tribondeau suggested in 1906 that radiosensitivity is greatest for undifferentiated cells and those undergoing active mitosis, and least for well-differentiated and nonreproducing cells; this generalization is fairly well substantiated in practice.⁸ Modern radiotherapy rests on both natural differential radiosensitivity and recovery rates. It is these *differences*, often referred to as the "therapeutic ratio," which make it possible to destroy a tumor while preserving enough of the tumor bed for patient survival with reasonable function and comfort. An enhancement of the therapeutic ratio is sought of chemical agents. Unfortunately most protective and potentiating chemicals are rather unspecific in their action and have so far proved of limited practical value.

Protective agents include steroids, sulfhydryl substances like cysteine and cysteamine, and hypoxia. Synthetic vitamin K and oxygen have been used as potentiators. Of all these agents, oxygen appears to be the most promising. We shall consider it specifically below.¹⁰

THE OXYGEN EFFECT. It is widely accepted that most biologic effects result from indirect chemical effects, involving radicals pro-

duced in an aqueous solution in cells. Oxygen in solution greatly affects production of the most active radicals so that oxygen tension strongly influences biologic response to electron and x-ray beams. It should be stressed, however, that heavily ionizing particle beams, like those of alpha rays, protons, and fast neutrons, act in the main *directly* on vulnerable cellular components. Consequently, they are much less dependent on the presence of oxygen and produce their effects even in anoxic tissue.

Since effective x-ray treatment depends on how much oxygen is in the tissues, any tumor hypoxia represents a serious limiting factor in x-ray therapy. Unfortunately, there is considerable direct evidence that something of the order of 1 percent of the cells in a solid growing tumor is hypoxic. This results from the tumor's outstripping its oxygen supply, both as a result of its rapid growth and capillary congestion and outright destruction. This situation is illustrated schematically in Figure 6.

Hypoxia makes tumor cells less vulnerable to x-ray therapy. Unfortunately, healthy tissues are not normally hypoxic; the radiation dosage, therefore, cannot be raised to levels adequate to kill the hypoxic tumor cells without simultaneously destroying the tumor bed.

It would appear from this theoretical argument that increasing oxygen supply to tumors should oxygenate hypoxic cells and render them more sensitive, thereby reducing the probability of tumor recur-

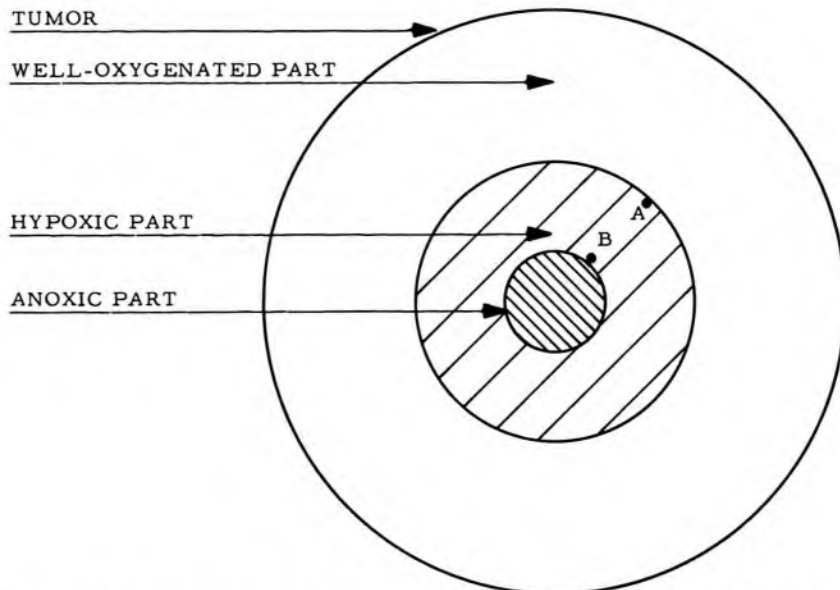


Fig. 6. Schematic representation of tumor hypoxia, showing three regions: well-oxygenated main mass of tumor; hypoxic transitional region—good O_2 supply at A, marginal at B; and necrotic center. Following irradiation, viable tumor cells may remain at B, protected from irradiation injury by their hypoxia.

rence. Gray conducted considerable research into this problem with cells both in culture and in vivo.¹¹ He found that the radiosensitivity of a wide variety of cell types changes through a range of about 2.5 to 1 as the oxygen concentration is varied. This work provides convincing evidence that increasing tissue oxygen tension can potentially increase tumor radiosensitivity. Figure 7 shows a general curve relating cell radiosensitivity with oxygen tension. For example, consider a hypoxic tumor cell versus a healthy cell (A versus B on the graph). The tumor cell is less than half as radiosensitive as it can become when fully oxygenated; the healthy cell is almost fully so. Assume we raise both cell oxygen tensions by the same amount. The tumor cell becomes more than twice as radiosensitive as before, while the healthy cell changes very little since it was already adequately oxygenated. The net result is a substantial improvement in the therapeutic ratio.

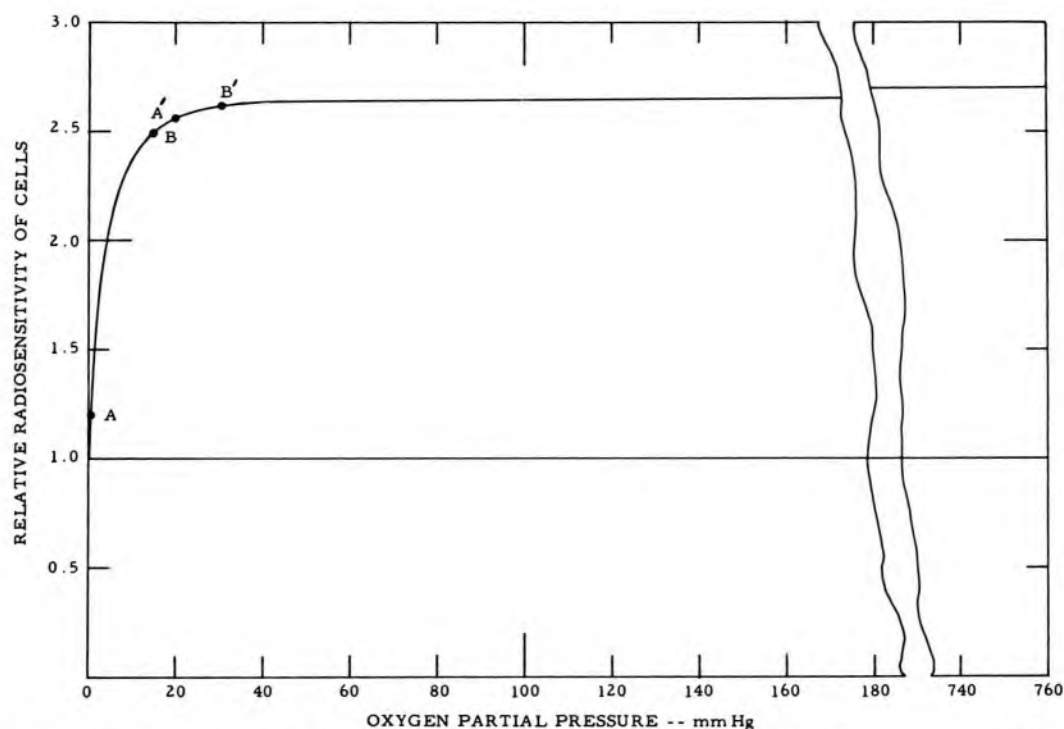


Fig. 7. The argument for increased x-ray therapy effectiveness at higher tissue oxygen concentration.¹¹ The curve itself. Tumor cells rapidly increase in tumor response when the oxygen concentration is increased from 0 to 40 mm Hg pressure (Gray). The argument. Healthy cells are already oxygenated, generally at 20 mm Hg and above (pt. B); hypoxic tumor cells are of course near 0 mm Hg (A). Hence, ordinary therapy cannot effectively sterilize the tumor cells without producing excessive damage to healthy cells. Increasing the oxygen supply raises the oxygen pressure of the entire system. We have assumed a 15 mm Hg O_2 increase for both, moving the points up the curve to B' and A'. Note that while initially the tumor cell was half as sensitive as the other, it is nearly as sensitive after oxygenation. (Actually, curves are not identical, but nevertheless quite similar, according to work of Gray and others.)

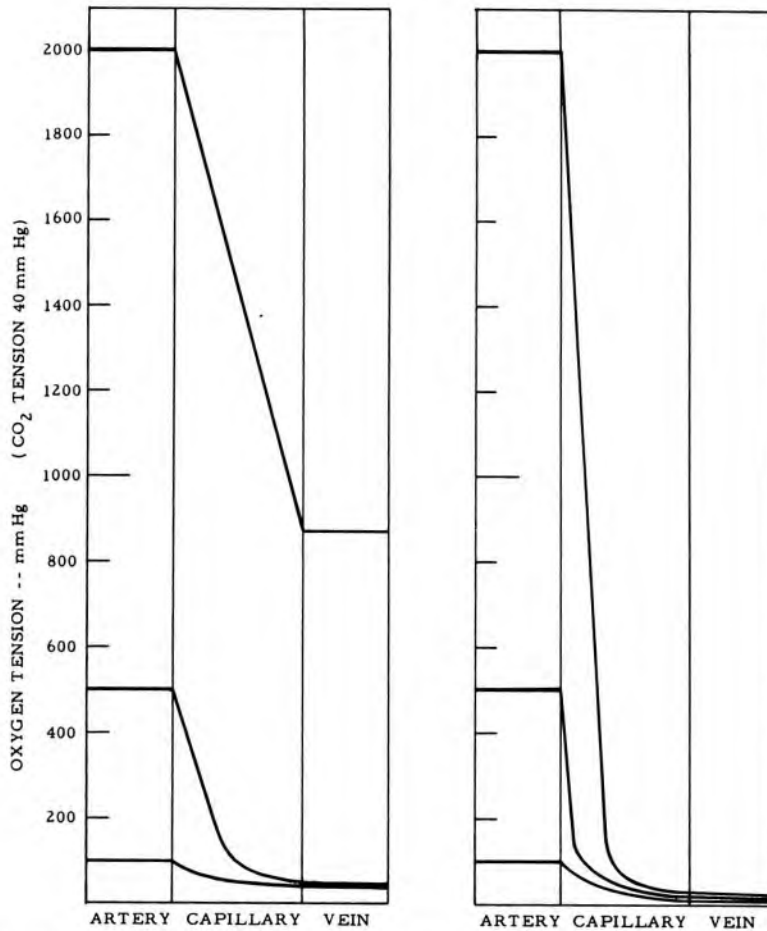


Fig. 8. Variation of oxygen tension in a capillary vs. arterial oxygen pressure.¹² Left. Normal capillary. Right. Congested capillary, 25 percent of normal flow. 100 mm Hg O₂ arterial pressure: ordinary air breathing; 500 mm Hg O₂ arterial pressure: pure O₂ at 1 atmosphere pressure; 2,000 mm Hg O₂ arterial pressure: pure O₂ at 3 atmospheres pressure.

An important theoretical point is that the oxygen must reach the cells themselves. Churchill-Davidson¹² has shown that high pressure oxygen can increase capillary and vein partial oxygen pressures substantially; significant increases are possible even when capillaries are congested (Fig. 8). This indicates oxygen can be brought to the region of the tumor and likely to the hypoxic cells as well. In practice, the patient is kept in a high pressure chamber breathing pure oxygen at three atmospheres pressure before and during radiation therapy, and later decompressed. Preliminary results indicate increased tumor response, encouraging cautious optimism about the ultimate benefit of hyperbaric oxygen radiation therapy.

An alternative method to increase tumor oxygen supply, proposed by Mallams,¹³ uses regional perfusion with hydrogen peroxide. Several theoretical questions are as yet unresolved concerning this tech-

nique, and the reported clinical results to date are too meager to permit more than tentative conclusions as to its effectiveness.

Rotblat has suggested approaching the problem from the opposite direction, or lowering the regional oxygen tension.¹⁴ This tends to equalize the radioresponsiveness of healthy and anoxic tumor cells because the former is reduced greatly, the latter very little. The obvious limitation to this procedure is that anoxia is particularly dangerous for already ill patients who are generally middle-aged or older. The technique has not yet been clinically evaluated.

Another possibility is to employ high LET radiation, which should exhibit much lower oxygen effect.¹⁵ This is very interesting theoretically, but the problem of obtaining sources of reasonable cost and availability has so far limited the investigation of this technique.

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6

Measurement of X-ray Dosage

In Chapter 5 it was shown that radiation effects depend on chemical and biologic as well as physical factors. However, the initiating event is always the radiologic lesion, which is determined mainly by the absorbed dose in rads to irradiated cells.* Consequently, accurate evaluation of the distribution of absorbed dose in tissues is basic to all treatment planning. This evaluation is by no means simple in either theory or application, so two full chapters will now be devoted to its coverage. This chapter deals with the units of radiation dosage and measurements made by physicists to provide basic data for the radiotherapist. Chapter 7 discusses dosage data and its clinical application more specifically.

TYPES OF X-RAY DOSAGE UNITS AND MEASUREMENTS

Before 1928 there was no unit of radiation "dose" at all. Therefore, radiotherapists were compelled to depend on exposure factors like kV, wave form, filter, mA, time, and treatment distance to reproduce treatments. Unfortunately, machine wave form and tube inherent filtration vary greatly among different machines, and even in the same machine from time to time. Moreover, kV and mA are not usually known very accurately, so reporting these factors cannot assure reproducible results among different machines. Holzkecht prepared crystals whose color changes when irradiated give an indication of total dosage, but these were of only limited usefulness. For many years skin erythema was used extensively as an indication of biologic dose. The erythema reaction is very variable, and is now believed to be of limited biologic significance; however, it did serve as a useful empirical guide as employed by some outstanding radiotherapists.

* At this writing, high LET radiation sources such as alpha, proton, and neutron beams are still in very limited clinical use.

The Roentgen (R)

After working on the problem for years, the International Commission on Radiologic Units in 1928 proposed the "roentgen"; the definition was revised slightly in 1937.^{1, 2} A unit of x-ray "exposure," the roentgen is essentially the amount of x- or gamma rays which can produce a particular amount of ionization per unit mass of air (1 esu per .001293g, or its equivalent, .000258 coulomb per kilogram). The roentgen is more fully discussed later in this chapter (p. 179). For many years the roentgen was felt to be an adequate unit and is still useful today. It was recognized even in 1928 that radiobiologic effects result from ionization in tissue. Air is a good substitute for tissue because its mass energy transfer coefficient $(\mu/\rho)_{en}$ varies similarly to that of water over a wide photon energy range, and its gaseous state facilitates ionization measurement.

More recently two basic limitations were recognized in the roentgen. First, many types of ionizing radiation consist of subatomic particles rather than photons, and the roentgen is defined only for photons. Second, the roentgen is not readily applicable to measurement of x-rays generated above 2 MV, so it is not useful in the linear accelerator and betatron x-ray region.

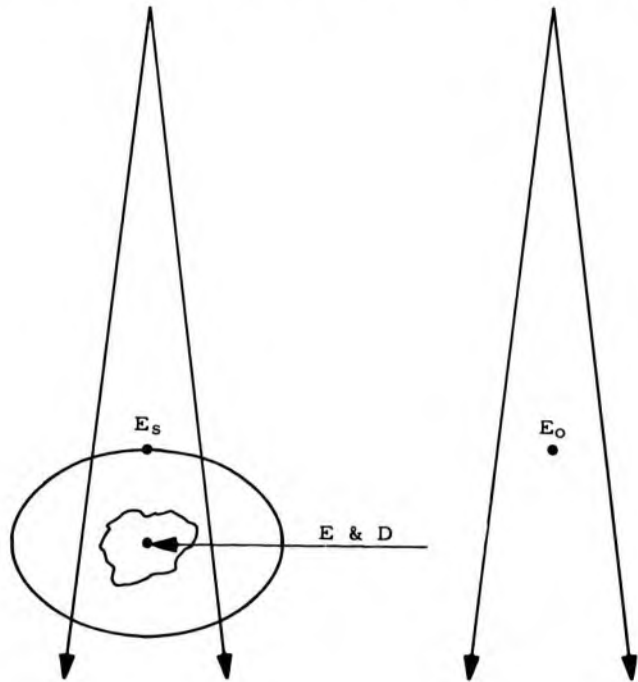
The Rad (r)

In 1953 the "rad" was proposed. It avoids the two major problems of the roentgen definition by concentrating on the *absorbed* energy rather than the incident beam itself. Applicable to any type of radiation, it is simply an absorbed dose of 100 ergs per gram of tissue (or, in general, any other absorbing material).³ As mentioned above, the rad gives a fairly good indication of the local radiologic lesion, for a low LET radiation like electron and photon beams.

The rad is used for all clinical work. However, for very high energy radiation and for nuclear particles a possibly more useful unit is the "kerma" (from *kinetic energy released in material*). It is similar to the rad, but includes not only the energy delivered *locally* but also *radiated energy* from bremsstrahlung produced by high energy particles.⁴ (Note this does *not* include scatter which is *deflected* but not *produced* by the absorber.) Bremsstrahlung radiation is negligible in ordinary clinical work, even supervoltage therapy, so the rad and kerma are usable interchangeably. However, the distinction becomes significant at high energies, especially for particle beams and high Z absorbers.

As we shall see, the roentgen and rad are both used in radiotherapy dosimetry.

Fig. 1. Irradiation of a patient for a deep tumor. A. Patient in place. E_s : skin exposure, roentgens; E : tumor exposure, roentgens; D : tumor absorbed dose, rads. Right. Patient removed. E_o : machine exposure output at skin distance, roentgens (air-R).



Types of Measurements

Now let us consider the types of measurements of importance to radiotherapy. Figure 1 represents a patient receiving external beam irradiation to treat a deep tumor. One desires to know D , the absorbed dose at any location of interest in and around the tumor. For photon beams of E_{eff} up to about 1 MeV, a common method of obtaining this value involves *three* steps:

1. Determine the machine x-ray output exposure in *roentgens* E_o during the treatments (Right).
2. Relate this to the resulting skin and depth exposures in roentgens— E_s and E in (Left).
3. Finally, convert the depth exposure E in *roentgens* to the absorbed dose D in *rads*, at any given location.

The first two steps involve direct measurement, the third the application of simple conversion factors. The x-ray machine output is measured, usually by a physicist, in roentgens per minute for all treatment factors on each therapy machine. The data relating depth dose to machine exposure are also initially measured by physicists, but under laboratory conditions. The first is called a measurement of “air-R,” since no tissue is involved. The second is called “depth dose” measurement. Depth dose data have been measured for most photon generators, treatment fields, and other variables, and accurate published information is generally available for clinical use. (See Chap. 7.) Depth dose is measured in *phantoms*, which are generally simplified rectangular arrays of tissue-equivalent material such as water. These

simulate the patient attenuation characteristics closely enough to provide useful approximations to the real patient dosage.

Exposure measurements on higher energy supervoltage beams are impractical for reasons discussed below, so a different procedure is needed. This usually employs direct measurement of the absorbed dose rate in a water phantom of standardized design, at a 5 or 10 cm depth, using the particular treatment machine of interest; the appropriate depth dose data are then applied to obtain depth dose values at other locations. The measurement must be repeated for all field sizes and beam qualities employed. This procedure also has potential advantages for lower energy beams and will likely be adopted in the future.

Ionization chambers are most generally employed to carry out depth dose and exposure measurements because they yield the most reliable data. Consequently, this chapter deals primarily with ionization chambers, their general nature and their use in measuring air-R and depth dose. However, other radiation detectors are occasionally used for specialized work, and we have therefore included them in a short section.

IONIZATION CHAMBERS

An ionization chamber consists of a volume of air or other gas with an associated pair of electrodes which can be placed at a location of interest. Ionization of the gas by radiation produces charges which are collected and measured, using a dc voltage across the electrodes and a suitable instrument. Table 1 presents a partial summary of gaseous detectors of ionizing radiation. If the system is properly designed and used, the collected charge provides a measure of the ionization per gram of the gas. This result can be used to determine exposure (roentgens) and absorbed dose (rads). Some units are designed for air-R measurement only and hence always read roentgens. Others are designed for depth dose measurements and can be used to measure by appropriate procedures both roentgen exposure at depth ("depth-R") or absorbed dose at depth in rads.

We discuss specific designs in the next sections. There are, however, two fundamental problems involved in using *all* ionization chambers. The first is how to relate the measured gas ionization per gram to the roentgen or rad dosage. The second is how to avoid certain pitfalls basic to all ionization chambers. These we will now consider.

Dosage versus Ion Density

AIR-R. Before discussing the more complex question of depth dose in a phantom we first consider the use of a "thimble" or "cavity"

TABLE 1. GASEOUS DETECTORS OF IONIZING RADIATION

Basic Type	Instrument	Application
Ionization chamber	1. Free-air chamber*	Primary standard of the roentgen
Ionization chamber	2. Cavity or thimble chamber a. condenser type (Victoreen)* b. ratemeter type c. tiny volume insert chambers (Sievert, British BD-2, etc.) d. experimental tiny chambers with cables to instruments	a. general air-R calibration b. air-R calibration and also isodose measurements c. dosage distributions around radium, in vivo and in phantoms d. depth dose measurement
Ionization chamber	3. Monitor chambers*	Mounted in tube housing and traversed by beam during therapy. Used to check machine output consistency.
Ionization chamber	4. Extrapolation chamber*	Provides best central ray percent depth dose and back-scatter data.
Ionization chamber	5. Cutie pie [†]	Portable radiation survey meter for x- and gamma rays, area safety monitoring.
Ionization chamber	6. Pocket ionization chambers [†]	Personnel radiation safety monitoring during hazardous procedures.
Ionizing-event counters	1. Proportional counters [‡] 2. Geiger-Mueller [‡] (G-M) counters	Radiation particle research. General radioactivity and neutron particle detection.

* Of major importance in x-ray work; considered in some detail (see text).

[†] These are considered separately in Chapter 13.

[‡] These are considered separately in Chapter 10.

chamber for measuring air-R. This type of ionization chamber has a thimble-shaped outer electrode with a thin cylindrical electrode in the center.

The electrodes introduce two basic sources of error (Fig. 2). First, the thimble wall both attenuates x-rays before they reach the thimble air and scatters photons *into* the air volume. In practice, errors

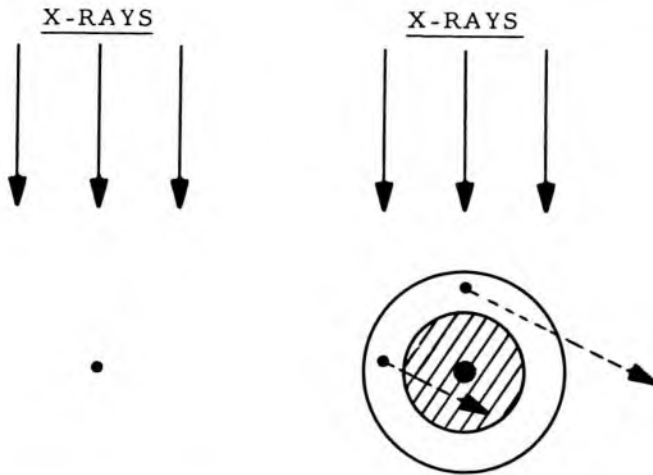


Fig. 2. Thimble chamber wall effects. Two effects exist. First, the x-ray intensity is reduced by true x-ray absorption (dot) and scatter (dotted lines) by the wall. Second, photoelectrons from the wall ionize the gas differently from an equal mass of a true "air-wall," material. Left. Free air situation. Right. Thimble situation.

from this cause can be minimized by design and calibration of the chamber for the particular kind of radiation employed. A second source of error is perhaps even more important. It results from the fact that the chamber wall must of necessity be of different chemical nature from the air volume where the ionization is measured. Both the chamber wall and the electrodes therefore contribute disproportionately greater or lesser amounts of photoelectrons than air, depending on the Z of the materials used. As a result, one finds a pronounced dependence of calibration on the thimble materials and beam spectrum.

In general, the energy dependence can be minimized by making the chamber wall out of material very similar to air in attenuation characteristics (such as carbon) and making this wall and other electrodes relatively small. In National Standards units, walls may be avoided altogether (p. 178).

Let us assume now that appropriate corrections can be made for this "wall effect." One can very easily infer roentgens from the measured ionization per gram, since a single numerical factor relates ions per gram to roentgens.

BRAGG-GRAY RELATIONSHIP AND DEPTH DOSAGE. When depth dose data are required, a more complex situation arises because of the dissimilarity of air and tissue. In 1936 Gray derived a basic relationship of primary importance to depth dose measurement. Based on some work of Bragg, the conclusions are usually called the Bragg-Gray Theory. Since it is so fundamental we shall briefly discuss the concepts involved.⁵

Consider a depth dose measurement arrangement in which phantom material (dashed lines) is replaced at depth by a tiny volume of gas at P , in Figure 3.

Our objective is to measure the ionization produced in a given mass of this gas (Right) and then to use some simple calculation to

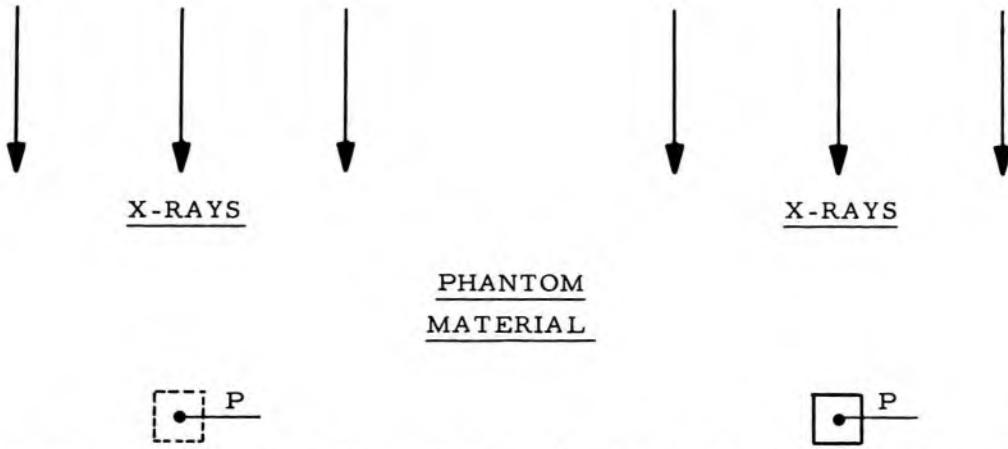


Fig. 3. The Bragg-Gray relationship. Left. Actual situation. X-rays strike phantom material at P. We desire the dose in rads for the tiny dotted volume around P (D). Right. Measured situation. The dotted volume of phantom material around P has been replaced with air. Then, the ion pairs produced per gram of air present are measured (J_m). The Bragg-Gray relationship tells how to obtain D from the measured air ion density J_m (see text).

infer the absorbed dose at the same point before the gas was put there (Left). The relationship is reasonably simple in form:

$$D = (0.01 J_m W) s_m \tag{6-1}$$

Here D is the dose that we seek in rads. J_m is the measured ionization per gram of air in the tiny volume—that is, the ion pairs per gram of air. But rads are expressed in *ergs*, not ion pairs, per gram. For this reason J_m must be multiplied by another quantity, namely W , which is the average energy in ergs needed to produce an ion pair. It should be stressed this factor W varies with both photon energy and the gas used. Then, $(J_m W)$ is the ergs per gram of air. One hundred of these would be one rad in air, by definition of the rad, so this explains the $(0.01 J_m W)$ expression.

The remaining item to be explained is the s_m term at the end. Up to now we have computed the absorbed dose in the *gas*. We actually want the absorbed dose in the *medium*. A natural question is why should there be any such difference between the gas and the medium when the same primary electrons are released in both? The answer is that different materials are ionized to different degrees by the same energy electrons striking them. As a result, an electron of a given energy produces a different number of ions in the *gas* than it would in traversing an equal mass of *tissue*. S_m expresses the ratio between these numbers and is called the “effective stopping power ratio” for the

medium. For elements like oxygen, nitrogen, and carbon, s_m is nearly unity compared with air. However, hydrogen is extremely atypical and has stopping power ratios of the order of 2.5.⁶ Since hydrogen is a major constituent of tissue, stopping power must be specifically considered in deriving rads directly from ionization measurements.

PRACTICAL ASPECTS. This theory is all basic* but fortunately need not be applied explicitly in practical dosage evaluation. When ionization chambers are used in phantoms, the usual procedure is to obtain the true roentgen dose and then compute absorbed dose from this, by using available appropriate tabular factors (Chap. 7). Two simple conditions must be met for the ionization chamber to accurately indicate roentgens at depth. First, the chamber must read roentgens accurately in air for the radiation quality present in the phantom; this would seem to be a reasonable requirement. Second, the chamber wall must be thick enough to absorb all primary electrons from the medium before they reach the air volume, assuring that only electrons generated in the chamber wall can ionize the gas and thereby simulate the situation in air.

Basic Problems in Use

Some problems can arise in *all* ionization chamber measurements. These relate to charge collection, determining the mass of gas ionized, and "electron equilibrium" effects.

CHARGE COLLECTION. In any ionization chamber, the number of gas ions produced by the radiation provides a measure of dosage. These ions must therefore be efficiently collected before they can recombine (or at least a known fraction), or there will be an indication of lower than actual dosage. Figure 4 shows a "voltage saturation curve" typical of ionization chambers. This is a graph of the current versus collecting voltage across the electrodes for a constant intensity x-ray beam.

At low collecting voltages (A), little ionization current is measured because slowly moving ions have time to recombine as they drift toward the collecting electrodes. With higher voltages these ions are swept up by the stronger electric fields and measured before they can recombine. In practice moderately large voltages assure good collection efficiency of the order of 90 percent or greater, and further increase in voltage does not produce much further increase in collected current. This relatively constant current operation (flat part of the

* The Bragg-Gray theory assumes a very tiny gas volume in a medium of essentially constant electron flux produced by the x-rays. Homogeneity of the medium, reasonable constancy of the x-ray intensity around the cavity, and electron equilibrium are assumed (see below).

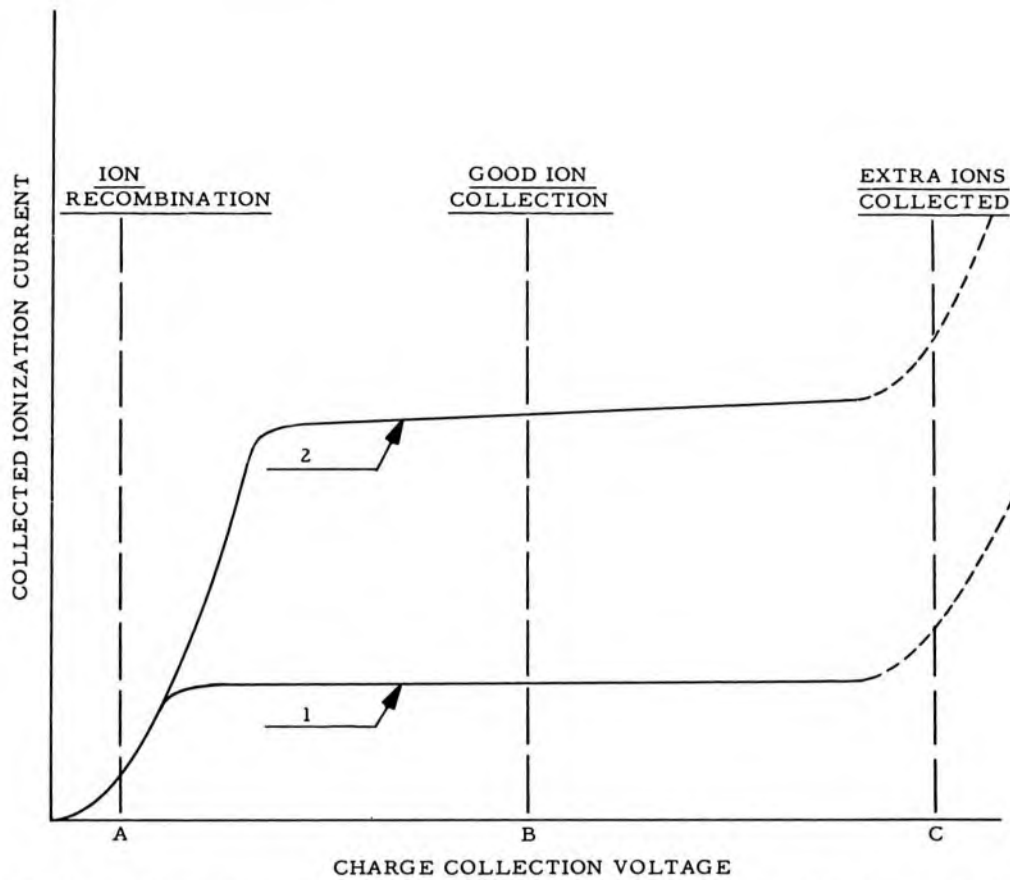


Fig. 4. Ionization chamber voltage saturation curve. A. Low collecting voltage: low current due to ion recombination. B. Saturation voltage: substantially complete collection of ions. C. Proportional region: disproportionately high measured currents due to production of extra ions by collision of original ones with air molecules. Note the flatness and length of the flat portion both decrease as the beam intensity increases (curve 2 vs. curve 1).

curve) is called "voltage saturation," and ionization chambers should be operated this way for maximum accuracy (B).

At higher beam intensities (curve 2), ions are produced more rapidly and hence occur closer together. They are, therefore, more likely to recombine, so a greater electric field is required to collect them efficiently. As a result, ionization chambers should always be operated at high enough voltage to assure saturation at the highest beam intensity measured.

The greater the collecting voltage, the more ionization current is measured. One cannot, however, increase chamber voltage indefinitely, for two reasons. First, insulation resistance is limited, and leakage currents increase with voltage. Since leakage gives a false indication of ionization current, it sets an upper limit on useful voltage. Second, too high electric field strengths may accelerate ions to velocities at

which ion multiplication occurs (see Chap. 10), invalidating the measurement (C).

It should be emphasized that when using an ionization chamber, one must provide an adequate collecting voltage for the particular beam intensity measured, to avoid serious recombination errors. This type of problem can arise when one uses instruments designed for low intensity work to measure high intensity beams. It is wise to verify that voltage saturation exists in any such unusual usage. This can be done by making a few measurements at a constant beam intensity with higher and lower collecting voltages.

MASS OF GAS. The roentgen definition specifies one esu of charge per .001293g of air, which is the mass of one ml of dry air at 760 mm of mercury pressure and 0°C. If the ionization chamber is sealed, the amount of gas is constant and does not change with pressure and temperature. On the other hand, if the chamber is not sealed, the air density becomes the same as that in the room in which the measurement is carried out. The mass of air in the chamber which can be ionized can then vary greatly, so the number of collected charges responds to changes in pressure and temperature. For example, consider a chamber calibrated at 760 mm Hg (at sea level). If this is taken to Denver, Colorado (a mile above sea level), it may read only about (600/760) or 0.8 times as high for the same roentgen exposure. (At this elevation the air is thinner, so it provides less molecules to ionize.) Correspondingly, using a Victoreen ionization chamber at 30°C instead of 20°C for which it is calibrated causes it to read 3 percent low. More generally, assume an unsealed ionization chamber reads correctly at a pressure P_0 and absolute temperature T_0 . At a different pressure P and temperature T , it will read D instead of the correct value D_0 . Then, the true value D_0 can be calculated as:

$$D_0 = (P_0/P) (T/T_0) D \quad (6-2)$$

ELECTRON EQUILIBRIUM—PRACTICAL EFFECTS. This concept arises in two important applications in radiology. The first is the "skin sparing effect" of supervoltage radiation. In supervoltage therapy the surface ionization density (surface rad dose) is a *minimum*, rising to a peak subcutaneously beyond which it decreases with depth due to photon absorption. The depth at which this maximum dose occurs increases with photon energy: depths for 2MV, 4MV, 8MV, and 22MV x-rays are, respectively, about 4, 10, 20, and 40 mm in soft tissue! For cobalt-60 teletherapy, as low as 30 or 40 percent of the maximum dose is observed on the skin, with a maximum dose 5 mm below the skin. This effect is of some clinical importance and is discussed further in Chapter 7.

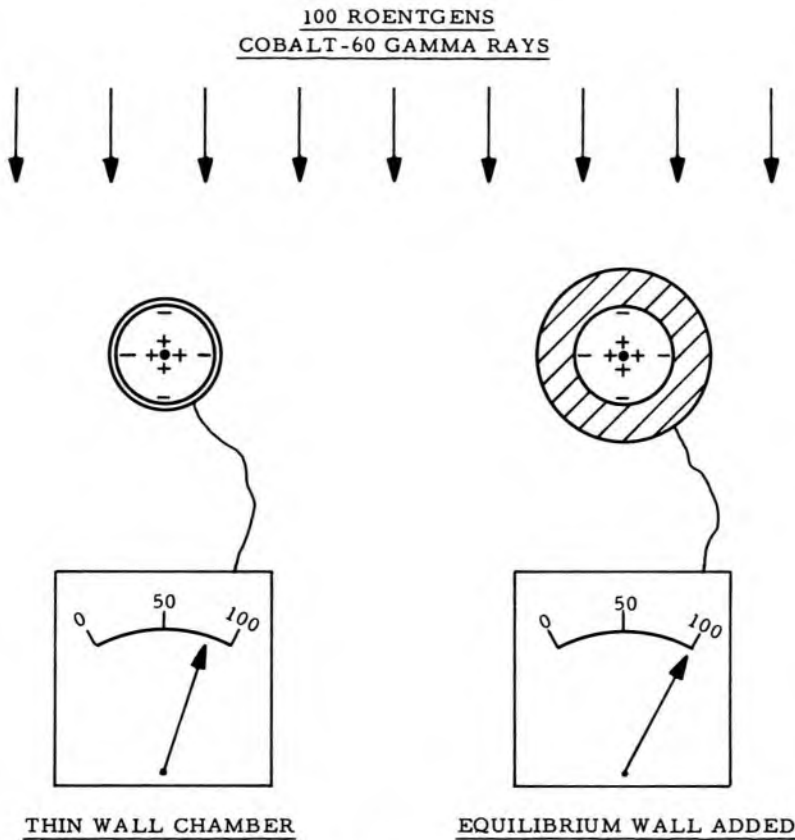


Fig. 5. Exposure of a thin walled ionization chamber to cobalt-60 rays. Left. No added cap; exposure reading is low. Right. Equilibrium cap added; full reading of exposure is obtained.

The second application is in measurement of roentgen dosage. If precautions are not observed ionization chambers can give quite *low* indications of dosage with supervoltage x-rays, yet read quite *accurately* in measuring 250 kV x-rays! For example, consider the thin-walled ionization chamber in Figure 5 (Left), exposed to 100R of cobalt-60 radiation. The indicator reads only 80R. Suppose we add a sleeve of plexiglas about 4 mm thick as an "equilibrium cap" and again expose the chamber to 100R as before. The indicator now reads the full 100R [Fig. 5(Right)]—just because extra material was placed in the beam! If one repeats the test with 250 kV x-rays, the two readings are nearly the same. (Actually, that at Right is slightly lower because of the added x-ray attenuation of the cap.)

ELECTRON EQUILIBRIUM—EXPLANATION. Figure 6(Top) shows a narrow beam of supervoltage x-rays striking a soft tissue surface. In well-filtered supervoltage beams, the photon energies range from 500 keV and upwards. Compton scatter is dominant, in fact almost exclusive for teletherapy* and 2 to 8MV x-rays. Referring to Table 3 in Chapter

* Teletherapy machines are units with very high radioactivities of cobalt-60 or cesium-137. Their gamma rays are used for supervoltage therapy.

National Standard Units

Free-air ionization chambers are used by national laboratories in many countries to measure exposure from x-rays generated between 5 kV and 2 MV. Almost all are of the parallel-plate type (Fig. 7), but Sweden's is a special cylindrical unit designed by Thoraesus. For x-rays generated at 500 kV to 2 MV and for gamma rays with energies up to those of cobalt-60, very large and high pressure free-air chambers are used, as well as special design graphite cavity chambers. Intercomparison of such instruments shows general agreement to well within ± 1 percent, and analysis indicates free-air chambers should be this accurate. At higher energies, the usefulness of the roentgen itself is doubtful, and dosimetry generally involves additional uncertainty.

Secondary standards for exposure rate meters are far less accurate, and ± 3 percent uncertainty exists in National Bureau of Standards calibrations of clinical dosimeters.⁷ Evidently one could easily measure 5 percent and greater apparent differences in x-ray output on the same stable x-ray machine using two different standardized clinical dosimeters!

FREE-AIR IONIZATION CHAMBER.⁸ The free-air ionization chamber is of great importance in defining the roentgen and merits discussion in some detail. It is indicated schematically in Figure 7. A very highly stabilized x-ray machine is employed at location S (left), and the beam

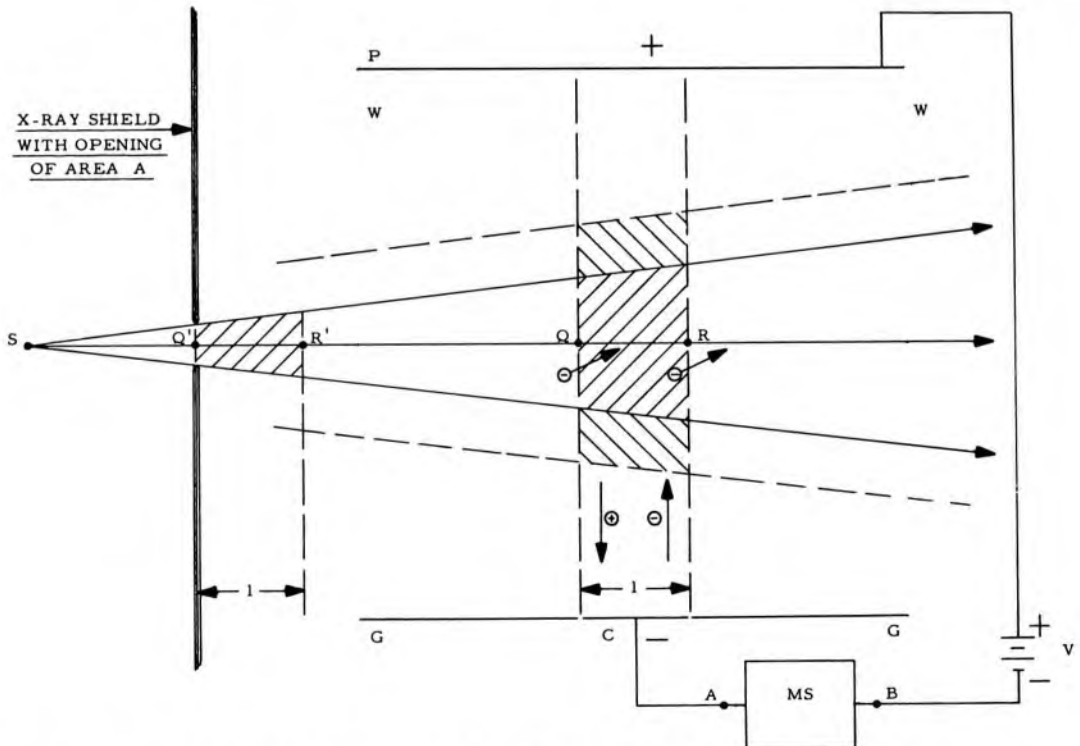


Fig. 7. Free-air ionization chamber—basic principles (see text).

is very well collimated. (Two collimating holes are sometimes employed to more precisely determine the air volume.) The beam emerges at Q' from the hole in a gold ring of area A, then traverses the air between the plates P above and G, C, and G below. The beam actually emerges without striking anything except some very fine aluminum wires W, placed approximately 1 cm apart, which run parallel to the upper and lower plates. Since the beam does not strike any electrodes or wall, the device is called a "free-air" ionization chamber. All of the plates are rectangular in shape. The top plate is connected to the positive terminal of an appropriate dc collection voltage V. The central bottom plate C collects charges which are measured by the special null detector measuring system MS. ("Null detector" means the device measures the charges without developing a significant voltage between points A and B. In practice this could be a high quality potentiometer system with an electrometer detector).

THE ROENTGEN. We shall now define exposure more formally than previously and explain what is needed to measure it accurately. Paraphrasing the most recent definition,⁹ exposure is the quotient: $\Delta Q/\Delta m$, where ΔQ is the sum of electrical charges on all ions of one charge produced when all electrons liberated by photons from an air volume of mass Δm are completely stopped in air. The special unit of exposure is the roentgen (R), which as stated previously is one esu per 0.001293 gram or its equivalent 0.000258 C/kg of air.

Many conceptual subtleties of this definition require specific comment.

1. *The sequence of events.* Photons strike the volume of mass Δm and liberate primary ordinary electrons ("negatrons") and, at higher energies, positrons also. These energetic primary electrons produce secondary ionization in both the original and surrounding volumes.
2. *The charges produced.* The total charge produced of one polarity (i.e., ordinary electrons) is of interest here; since materials are ordinarily electrically neutral, an equal number of positive ions also results. The definition adds an important stipulation: these charges must all be produced *in air*. The reason is that more or fewer ions are produced by the same primary electrons traversing different materials.
3. *The mass of air.* 0.001293 gram is the mass of one milliliter of dry air at NTP (0°C, 760 mm Hg pressure). Consequently, a milliliter of ordinary air may differ from this both in being moist and at some other temperature and pressure. Correction for air moisture is usually unnecessary in clinical work; for example, saturated air at 30°C contains only about 4 percent

moisture by volume, which introduces less than one percent error in a reading. However, as indicated above, pressure and temperature corrections *are* generally necessary (p. 174).

4. *The $\Delta Q/\Delta m$ concept.* The air volume must be reasonably small to properly sample the region of interest. For example, the intensity of a cobalt-60 teletherapy beam is 4.5 percent less at 82 than at 80 cm from the source. Consequently, a 5 cm diameter ionization chamber would yield an inaccurate average figure if used in this application. On the other hand, one cannot theoretically decrease the chamber size excessively (below 1/20 mm on a side) because substantial statistical uncertainties then arise from the discontinuous nature of ionization processes. Electrical considerations usually require that ion chamber dimensions be of the order of a few millimeters at least, so the limitations in size are practical rather than theoretical in most applications.

In summary, measurement of exposure requires the following: a means to select a known volume of air (at known pressure and temperature) whose ions can be uniquely ascribed to that volume when it is irradiated, and means to collect and accurately measure the charge produced. When these requirements are both met, the number of roentgens is numerically given by the collected charge in esu divided by the air volume in milliliters, corrected, when indicated, for pressure and temperature. Hence, one must essentially measure a charge and a volume.

Let us now return to Figure 7, to see how these requirements are met in the free-air ionization chamber.

CHARGE COLLECTION. Any ion pairs produced between P and C and the heavy dashed lines tend to be collected by the voltage V, as indicated by the vertical arrows. The vertical dashed lines represent the limiting paths of positive ions traveling from P to C; they are made almost perfectly parallel by the use of three means: the plates G, the thin aluminum wires W, and the use of a null detector. The guard plates cannot be made too large since the unit must be reasonably portable, so the wire array supplements their action. This is accomplished by spacing wires evenly both in position and voltage.

Of course, as in all ionization chambers, reasonable voltage saturation must be assured; because of the precision required here, ion recombination corrections are also usually made.

A comment is in order about the separation of the plates and the materials of which they are made. While the geometric margin of the beam is indicated by the solid lines diverging from S, some energetic electrons originating within this volume are driven or deflected beyond

the beam margin, as far out as the dashed line some distance on all sides from the margin of the beam. This distance increases with the maximum range of a recoil electron at the photon energy employed, and at very high energies may become quite significant. This is one of the reasons why these chambers cannot be used with as great accuracy above about 500 keV effective energy. The definition of the roentgen requires that none of the primary electrons may strike any material other than *air*. For this reason also, the plates must have an adequate separation so that very few primary or other fast electrons can strike them.

The plates are normally made of a relatively low atomic number material, such as aluminum, because some photons are scattered out of the beam and strike the plates. Such photons will be relatively few in number because primary beam photons are confined to air, which does not scatter very effectively. A few nevertheless strike the plates and release electrons. If high atomic number materials are used, photoelectric emission from the plates can contribute significantly to the measured ionization current. The use of aluminum throughout the chamber minimizes this effect.

THE AIR VOLUME. This "air volume" is difficult to define in the context of the use of this ionization chamber because the beam diverges. We measure charges produced between the indicated vertical dotted lines and in the shaded volume. However, the intensity of the beam is different at location Q from that at R because the beam diverges. To be precise, one would have to make the length "l" relatively short compared with the total distance to the source so that the intensity is about the same at Q and R. This, however, would sharply reduce the collected charge.

To overcome this experimental dilemma two expedients have been employed. First, advantage is taken of the fact that most x-ray photons are not significantly absorbed in air. For example in Figure 7, of 3 billion photons traversing opening Q' substantially 3 billion photons impinge upon the shaded volume at Q. Of course, with very soft radiation some photons are absorbed along the way; this is normally not excessive, however, and is easily corrected for in practice. The second expedient is to say that the total air ionization along a fixed length of beam is independent of whether the beam is diverging or parallel. This assumption is justified because as divergence *reduces* the beam intensity it correspondingly *increases* the beam area. If there is negligible attenuation the compensation is almost perfect.

For these reasons we can say that the primary ionization produced in the shaded volume to the right of Q is, with appropriate allowance for air attenuation, identical to the ionization produced in a

similar length truncated cone $Q'R'$. Furthermore, this in turn is the same ionization that would have been produced had the beam been a parallel, rather than a diverging, beam. Consequently, the charge collected by the collecting plate is the same as would have been produced by primary ionization in a volume (Al) at location Q' . *Note that even though we measured the charge at some distance from the opening A, the charge and volume involved are all appropriate to the location of the diaphragm, and not to the location of actual charge collection.*

Consider a numerical example. Given: $A = 2\text{cm}^2$, $l = 5\text{ cm}$, and 400 esu are collected in 2 minutes. What is the exposure rate at the diaphragm in roentgens per minute? Intensity in

$$R/\text{min} = \frac{\# \text{ esu}/\text{cm}^3}{\# \text{ min}} = \frac{(400)/(10) \text{ R}}{2 \text{ min}} = 20 \text{ R}/\text{min}.$$

OTHER ASPECTS. As in any other ionization chamber the free-air chamber reads low values in measuring supervoltage x-rays unless electron equilibrium exists. In Figure 7 some primary electrons from the shaded volume QR leave it and produce ions to the right of R; these ions are not collected, so a low indicated dose can result if the loss is not replaced. However, some electrons produced to the *left* of Q cross into the collection volume and contribute ions. Equilibrium exists if these opposing tendencies compensate. This is achieved when enough air or other material exists between S and Q to provide sufficient primary electrons; if this requirement is not met, the measured ionization will be *low*. Substantial distances SQ are required at atmospheric pressure to accurately measure cobalt-60 photons, so high pressure chambers have been employed. (This design of chambers also reduces the required plate separation.)

Finally, once the esu per cm^3 is measured an air-density correction must be made. Equation (6-2) can be used for this purpose. Here $P_0 = 760 \text{ mm Hg}$, $T_0 = 273.12^\circ\text{K}$.

National laboratories have developed special instruments, which are essentially thimble chambers of more sophisticated design, for intercomparison of high energy standardizations. These yield accurate indications of exposure for high photon energies. They are transported between laboratories and their indications used to intercompare the basic standards. The interested reader is referred to the N.B.S. Handbook 85 for further description of such devices.¹⁰

Clinical Air-R Instruments

Clinical air-R measurements are generally made using cavity ionization chambers. Several types exist, each with its own advantages

and limitations. In this section we briefly indicate the basic types, their uses, and accuracy limitations. Because of its extensive clinical use we shall discuss the Glasser-Seitz condenser chamber instrument (the commercial Victoreen dosimeter) at some length.

BASIC TYPES. Three basic designs of cavity chamber x-ray detectors are indicated in Figure 8.¹¹ All have essentially the same chamber parts: an electrically conductive thimble surrounding a central electrode. The differences arise in the termination at the open end. In the top design a removable cap is used to protect the insulator from dirt and moisture and to help determine the gas volume. This type is useful when a relatively small chamber is required, such as in radium measurement and other applications where the dosage varies rapidly with position. Some small chambers have also been designed for depth dose measurement as well. In the center design a rigid stem contains a condenser which extends the measured roentgen levels to higher values. This is the Victoreen instrument type. Both chambers (Top and Center) are charged to a reference voltage using a special associated instrument. Some of this charge is neutralized by roentgen ex-

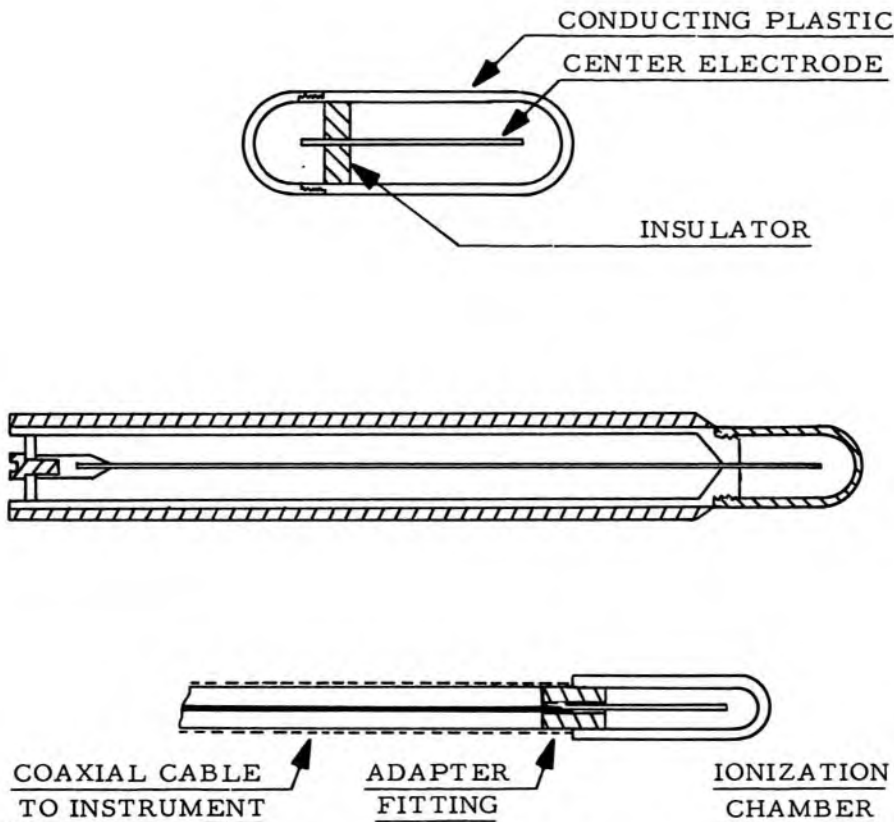


Fig. 8. Three types of cavity chambers. Top. Small chamber without cable or condenser (cap at left unscrewed to charge and read chamber). Center. Condenser chamber (see Figure 9 for details of construction). Bottom. Chamber with cable to instrument.

posure; the loss of charge is then measured on the instrument and the roentgen exposure thereby inferred (see below). This is called an "integrating" type of measurement. In the bottom design the thimble is located at the end of a flexible cable and the measured ionization current used to obtain an indication of roentgen exposure rate on an associated "rate meter." Unlike the other units, the cable type gives an immediate indication of output rate in roentgens per minute. This can be a great convenience. The two types of instrument indications are analogous to those of speedometer dials of an automobile. The speed indicator is a rate meter, indicating miles or kilometers per hour. The *total* traversed miles is shown on a counter called an "odometer" below the speed indicator. This gives a total, or integral indication of the distance traveled.

The Philips and Farmer-Baldwin dosimeter instruments are quite versatile in permitting both rate and integrating operation and can be used with cable or non-cable type ionization chambers. They are preferred as laboratory instruments. The Victoreen unit is less versatile but has the advantage of ready portability and is specifically designed for field x-ray machine roentgen output calibration.

Some "monitor" instruments are also used. These are generally permanently mounted in the x-ray beam to check the constancy of machine output in daily use.

ACCURACY ASPECTS. All of the previously discussed considerations of ionization chambers apply to these instruments as well. For example voltage saturation must be maintained, or they all read too low. In the integrating chambers of Figure 8 (Top and Center), the collecting voltage is supplied by the stored charge on the associated capacitance. (This charge is small in Top but much larger in Center which has a special condenser in the stem). Under irradiation this charge is partially neutralized, so the collecting voltage is correspondingly reduced. One must therefore be sure voltage saturation exists over the full range of roentgen exposure to be measured.

Cavity chambers are generally unsealed, so pressure and temperature corrections must be made. Also, their thimble walls and center electrodes are *not* generally air equivalent. This means calibrations vary significantly with beam quality, especially in measuring superficial therapy and diagnostic x-ray beams. Finally, one must be sure electron equilibrium is achieved in measuring supervoltage beams, for any type of ionization chamber. In addition, "stem effect" errors arise in measuring supervoltage x-rays (see below).

Cavity chambers also exhibit a distinct directionality effect. For example the Victoreen units are calibrated with the beam directed perpendicularly to the collecting electrode. If radiation is directed any

other way different sensitivities result. Some cavity chambers are intended for depth dose measurement; such units are generally designed specially to minimize directionality. As a rule, however, commercial cavity chambers are designed for air-R measurement only, and for the beam oriented perpendicularly.

It must be stressed that, in general, cavity chambers are *not* designed for depth dose measurements. Their use for this purpose can introduce substantial errors, both from directional and nontissue equivalence errors.

Condenser Ionization Chamber Instruments

In this section we shall discuss the Glasser-Seitz instrument, manufactured commercially by the Victoreen Instrument Company. Besides being one of the most commonly used calibration instruments in the United States, this type of dosimeter serves to illustrate the basic principles described generally above. We first describe the parts and operation of the instrument and then discuss pitfalls in its application.

DESCRIPTION.¹² The Victoreen instrument consists of two parts, the condenser chamber and the instrument proper (Fig. 9). The condenser chamber is a long, narrow tube with a thimble at one end. A cap (not shown) fits over the other end to protect insulator E from dirt and grease; in addition, it reduces the production and collection of stray ions at this end, which tend to cause a spuriously high dose indication. The instrument proper (not shown) provides a voltage supply to charge the condenser and an electroscopes-microscope combination to read the voltage on the condenser.

Figure 9 (Top) shows an axial cross-section diagram of the condenser chamber. At the extreme right is the thimble cap (D), usually nylon, screwed on to the rest of the assembly and enclosing the ionization chamber air volume (shown shaded). An aluminum center wire (C) is attached to the rear screw; it protrudes into the air volume and constitutes one electrode of the ionization chamber. The other electrode consists of a carbon coating on the inside of the thimble. This coating communicates through the threads of the thimble to the outer side of condenser assembly (B).*

The entire unit to the left of the dashed lines is the condenser part. A condenser is simply two conductors separated by an insulator of some kind. When equal amounts of charge of opposite sign are added to these two conductors, the condenser is said to be *charged*. The voltage between the two conductors depends on the amount of

* Caps may also be made of newer conductive plastics in which the carbon is permanently embedded in the volume of the plastic.

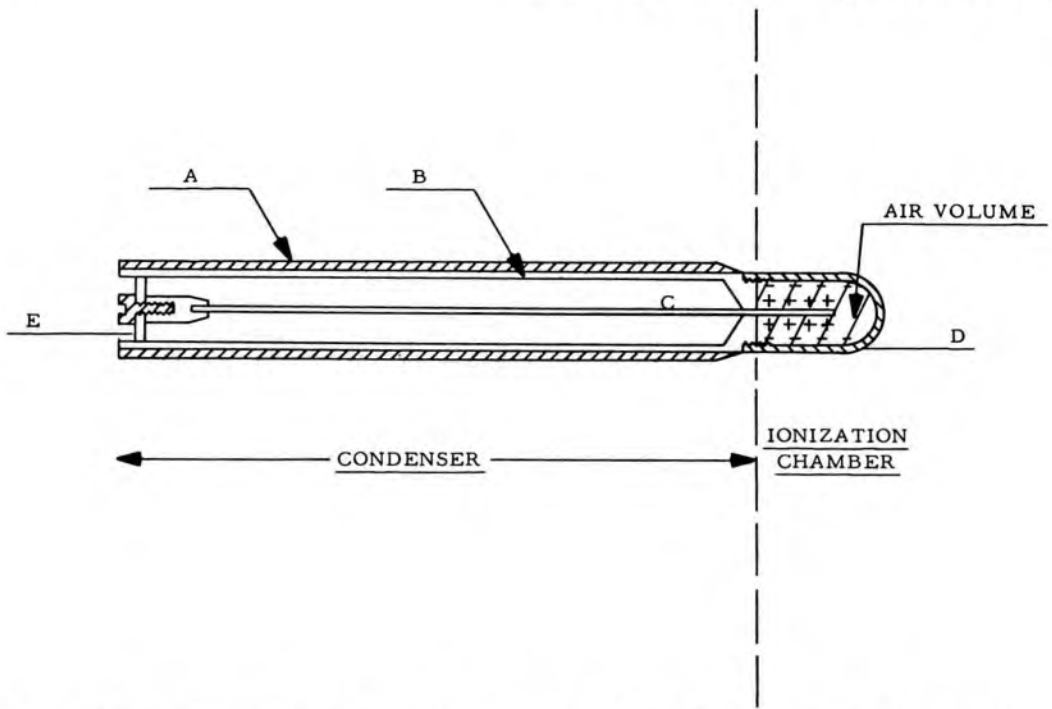


Fig. 9. Basic principle of condenser chamber instrument. 1. Condenser chamber unit. The ionization chamber (D) is to the right, the condenser to the left of dotted line. The chamber consists of thimble D with an aluminum wire C protruding into the air volume. The wire is the (+); a carbon coating inside the thimble is the (-) electrode. B is the polystyrene capacitor, or condenser. It supports the center wire C at the right end. B consists of a sleeve of polystyrene, coated with conducting graphite on both sides. The inner surface (+) makes contact with wire C by a spring (not shown). The outer surface (-) makes contact with both the protective sleeve (A) and the inner surface of the cap D. For charging B, contact is made to sleeve (A) and the screw electrically connected at the end of wire C. In use, a brass cap fits over the left end of the unit (not shown). This protects insulator E from dirt as well as reducing undesired charge collection at this end.

charge present: the more charge the more voltage. Should any of the charge be lost, the voltage falls; should the charge be increased, the voltage rises correspondingly. In this unit the condenser consists of a polystyrene annular or tube assembly. Polystyrene is employed because it is an excellent insulator. The unit is made by drilling out the center part of a rod. A final small hole is drilled in the center through which the aluminum wire is inserted. The inner and outer surfaces of this unit are coated with colloidal graphite, which is a fair conductor of electricity. As a result, the inner surface of this condenser consists of colloidal graphite which in use is charged positively, the outer surface of colloidal graphite which is charged negatively. The desired capacitance is achieved by control of the polystyrene wall thickness. To protect this relatively flimsy plastic capacitor, a brass sheath is employed which is at the potential of the outer electrode of the capacitor. The center aluminum wire is at the same voltage as the

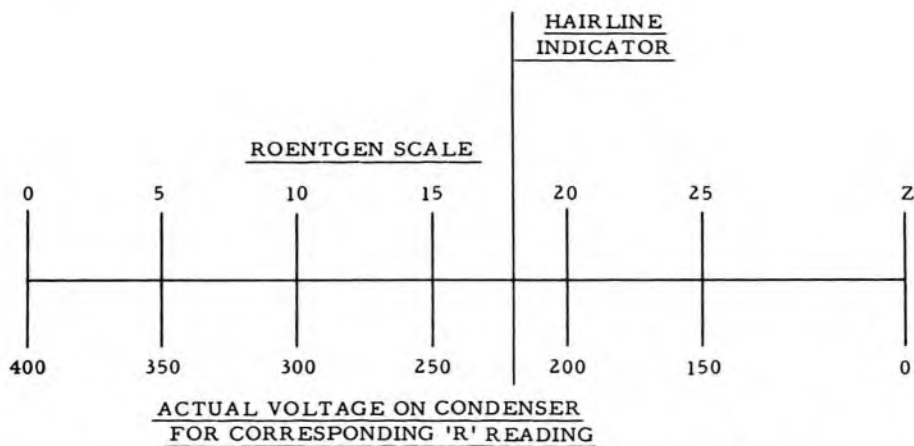


Fig. 9 (cont.). 2. The charger and electrometer unit. The instrument provides an adjustable voltage supply to charge the condenser and a string electrometer to read the condenser voltage when charged and after exposure to radiation. The figure shows the general appearance of the electrometer scale, with approximate condenser voltages corresponding to various roentgen indications of a 25 R chamber for the older Victoreen Model 70 instrument. For example, the reading shown is 18 R, with 220V on the condenser.

inner electrode of the condenser because of a spring contact between them (not shown). This center aluminum wire terminates in a screw and clamp at the rear, where it is supported by a polystyrene disc (E).

OPERATION. In operation the condenser chamber is first plugged into the instrument proper. This unit provides a voltage supply to charge the condenser, and a string electrometer and microscope unit to measure the resulting voltage, both during initial charging and after the chamber has been exposed to radiation. The dosage scale is indicated in Figure 9 (2).

Before charging, the instrument is checked with zero voltage to make certain the electrometer string is properly set mechanically. This is the so-called Z position. As indicated in Figure 9 (2), when there is no voltage across the condenser and the instrument, the string is on the Z line, indicating complete discharge. When voltage is now impressed on the system, charges flow onto the condenser chamber, and its voltage rises. This voltage also deflects the electroscopie hairline which moves to the left, reading zero on the scale (in roentgens) for the appropriate voltage. The condenser is then fully charged to the reference potential and charge and ready to use for measuring radiation. The approximate design value voltage for this condition is indicated below the drawing as are the corresponding readings for various stages of discharge of the chamber (for a particular model instrument).

Should the chamber now be removed and then reinserted without loss of charge, the string electrometer should again read zero R. On the other hand, if the condenser chamber leaks charge, reinsertion will give a false indication of dosage. These condenser chambers are initially quite high quality instruments and normally leak less than the equivalent of a small scale division in 24 hours (2 percent of full scale indication). If the leakage is greater than this, steps should be taken to correct the condition. If carefully blowing the rear end of the condenser chamber with air from a clean syringe does not accomplish this, the chamber must be returned to the factory for repair. (Solvents cannot be used on the rear insulator, as they generally attack polystyrene.)

Let us start with the condenser fully charged and the instrument reading zero roentgens. Now the chamber is inserted in the x-ray beam at the desired location, and the machine operated for an appropriate length of time, which is accurately measured and recorded. As a result of chamber irradiation during this time, ion pairs are produced in the air volume. Subjected to a significant collecting voltage field, substantially all of these charges are collected by the electrodes and serve to partially discharge the condenser.

If the condenser chamber-instrument combination is perfectly accurate, exposure of a 25 R thimble chamber to 15 R results in reduction of the condenser voltage from 400 to 250 volts. Therefore, when the condenser chamber is reinserted the new reading should not be 0 R but 15 R, corresponding to the new condenser voltage. The more roentgens received by the air volume, the more the condenser charge is neutralized, and therefore the lower the final voltage. Of course, even full scale (25 R in this case) cannot represent complete discharge because some collecting voltage is always required to prevent recombination of the ions in the air volume. For this reason, at least 150 volts remain, even for full scale reading of 25 R on the instrument. (In newer instruments somewhat higher voltages are employed than those indicated in Figure 9(2); the basic principle, however, is still the same.)

The above description applies to a chamber calibrated for 25 roentgens full scale exposure. In practice, 25 R true exposure will always result in a slightly different reading than 25 R on the scale because no instrument reads perfectly. The order of magnitude of corrections ranges from a few percent up to 15 or 20 percent depending on the particular instrument and photon energy employed.

TYPES OF CHAMBERS. Since the range of dosimetry applications is very great, many kinds of thimble chambers are provided by manu-

facturers of this type of instrument. The common chambers are suitable for beam qualities from 1 mm Al through 4 mm Cu HVL. For measuring cobalt-60 radiation it is desirable to buy a thimble chamber specifically designed for this work rather than attempting to use chambers designed for the orthovoltage range. Similarly, for very soft radiation such as that produced by beryllium window x-ray tubes, special very thin "end-window" ionization chambers have been developed.

We have shown a 25 R range in Figure 9(2). Condenser chambers with ranges from 0.25 R up to 250 R are routinely provided in the Victoreen instruments, and for some applications other ranges are available. However, all the *condensers* are the same, regardless of the associated *ionization chamber*. To change the sensitivity the manufacturer simply changes the air volume and other features of chamber design. For example suppose one desires an ionization chamber of 0 to 100 R range rather than a 0 to 25 R. For full scale deflection on the instrument we now must reduce the condenser voltage from 400 volts to 150 volts when 100 R, rather than 25 R, has been delivered to the air volume. This means that we require the same charge collection with 100 R as previously with 25 R. We can achieve this by reducing the volume approximately four times so that only one-fourth as many charges are produced for each roentgen as before. Thus, higher dosage ranges are achieved by using *smaller* air volumes, low ranges by using *larger* air volumes. The most common ranges for clinical work of these instruments are 25, 100, and 250 roentgens for measuring output of x-ray machines; and 0.25 roentgens for some other measurements. Many special ionization chambers have also been designed and are available, but these should be applied with full knowledge of their design limitations and inaccuracies, usually described by the manufacturers.

PROCEDURAL PITFALLS.¹³ Although there is nothing essentially mysterious about calibrating the output of x-ray machines, some pitfalls exist which can introduce major errors. Those relating to procedures include positioning, timing, and chamber cap replacement.

The ionization chamber must be positioned so that the central ray passes perpendicularly through the center of the air volume, which in turn must be the proper distance from the target. It is easy to position the chamber at the desired distance when adjustable collimators are used; however, cones usually get in the way, and one must generally set the chamber beyond the end. When *open* cones are used, a simple inverse square law correction may be made for the increased distance. To simplify arithmetic, the writer usually positions the ionization

chamber exactly one centimeter beyond the end of the cone. The correction factors for 50 cm SSD then becomes $(51/50)^2 = 1.040$, and that for 20 cm SSD is $(21/20)^2 = 1.102$. Of course, similar distance correction factors are easily provided for other SSD values.

Slight scatter errors arise when one calibrates *covered* cones. In orthovoltage beams, scatter from the cover may introduce errors of a few percent in calibration of x-ray machine output. A physicist can make appropriate corrections if desired for this effect, using an extrapolation technique.

Timing of the calibration exposure may introduce a second type of procedural error. Recall that intensity measurement with an integrating type instrument involves measuring the roentgen exposure obtained during an accurately timed period. For short exposures accurate timing is difficult. This problem is minimized by always using a chamber requiring a longer exposure time, such as the 250 R instead of a 25 R or 100 R chamber. In practice one tries to keep the exposure time in the range of 30 to 90 seconds.

In low voltage superficial machines there is usually less timing error because the timer and full kV x-ray exposure usually start and stop at almost exactly the same time (unless the timer is defective). However, in orthovoltage and supervoltage machines the full kilovoltage is applied to the x-ray tube gradually (Chap. 2). A period of up to several seconds may be involved during which the voltage rises smoothly or in steps to the final maximum value. During this "build-up time" the ionization chamber is irradiated at a lower exposure rate than the final value. As a result, the grossly computed roentgen rate is lower than the final value.

In some x-ray units a very rapid shutter [Fig. 10(B)] is provided so that this error is negligible. In other units [Fig. 10(C)] the shutter consists of two parts which separate in opposite directions as indicated. If the motor drive is fast enough in the latter case, the timer indication is very accurate because the ionization chamber is irradiated almost immediately upon pushing the button. However, in other machines either a rather slow or no shutter at all may be present [Fig. 10(D) and (A)]. This is not serious in treatment, when times are of the order of several minutes and the buildup time is only a matter of seconds. The calibration time, however, is only of the order of a minute, so buildup must be considered. A similar problem exists in certain cobalt-60 machines, in which a period of several seconds is required while a source wheel rotates into the irradiating position.

For such situations an appropriate correction for the timer indication is applied. This is evaluated most simply by making two successive exposures with different timer settings and extrapolating the graph of dose versus time to zero dose.

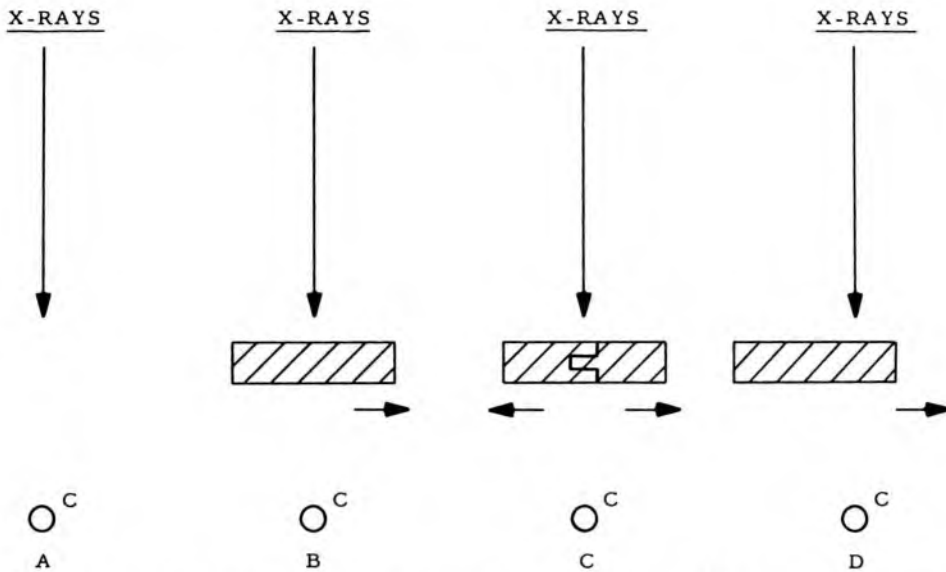


Fig. 10. Various shutter arrangements in x-ray machines; timing errors of exposure to chambers C below shutters. A. No shutter at all: *all* superficial units, *some* orthovoltage units, *all* supervoltage units. B. Fast shutter: moderate timing error. C. Split shutter: minimal timing error, usually. D. Slower shutter: timing correction needed.

A third procedural error of forgetting to place the cap over the rear end of the condenser chamber is occasionally made by inexperienced users of Victoreen instruments. These caps must *always* be employed, not only to assure that dirt does not destroy the accuracy of the instrument but also to keep out extraneous ions from the rear end of the chamber during irradiation. Reference to Figure 9 and a moment's thought will show that the center electrode at the rear of the condenser is at positive voltage with respect to the brass sleeve. As a result, ions can discharge the condenser at the *rear* as well as the *cap* end. In other words an auxiliary ionization chamber can exist at the rear of the condenser. Capping the chamber is usually enough to control this effect. When the measured field is so large that the primary beam overlaps the rear of the condenser, lead shielding of this part can be carried out but is usually not required with orthovoltage beams.

CORRECTIONS APPLIED TO DOSAGE READINGS. As with all unsealed ionization chambers, a pressure and temperature correction is required with Victoreen type chambers also. They are calibrated for 20°C, 760 mm Hg pressure, so $T_0 = 293.12^\circ\text{K}$ in formula (6-2). Should the chamber ever become sealed, an error can actually be introduced by this procedure, but this rarely occurs.

All these chambers have a substantial dependence on photon energy. For example, suppose that a given ionization chamber gave a reading exactly correct for an x-ray beam of 1 mm Cu HVL. This

chamber might be considerably in error for a harder x-ray beam, i.e., 2 mm Cu HVL or cobalt-60 radiation. Furthermore, it would most likely read too low in the diagnostic x-ray region and also much too low in the superficial therapy region (1 mm Al HVL). The range of variation in sensitivity of such chambers may be of the order of 15 to 20 percent. For this reason thimble chambers with their associated instruments should be calibrated for at least four qualities of x-rays and gamma rays: cobalt-60, hard orthovoltage, 4 mm Al HVL, and 1 mm Al HVL. For superficial and softer radiation, chambers specially designed for such work should be used, with their own factors. The resulting calibration factors are appropriate for the chamber and instrument combination used. If *either* is changed, a new correction factor is required.

When possible two or more instruments should be purchased. One should be maintained as a secondary standard, set aside in a safe dry place and employed simply for calibrating the other units. It should be sent to the National Bureau of Standards regularly for calibration at several energies. The other instruments should be employed for regular calibration and can be quite accurate if they are used carefully and intercompared regularly with the reference instrument. Calibrations may be altered when instruments are subjected to vibration or other trauma. Ionization chambers are particularly susceptible to trauma of a mechanical nature as well as electrical leakage from contamination of insulating surfaces. In particular, Victoreen thimble chambers should preferably be stored in a sealed container with fresh silica gel dessicant to maintain low leakage.

SUPERVOLTAGE BEAM MEASUREMENT. Ordinary orthovoltage ionization chambers are not suitable for supervoltage work without special thimble chamber caps to assure equilibrium thickness. Figure 11 shows the response of a 25 R chamber irradiated with four different quality x-ray beams of constant intensity. In each case the actual reading is compared with the maximum reading obtained for various thicknesses of unit density material added around the chamber air volume. Note that whereas orthovoltage radiation requires no additional cap to yield equilibrium readings, the other beams require a total of about 3, 15, and 40 millimeters of unit density material (including the chamber wall) for 2, 6, and 22 MV x-rays.

In cobalt-60 and 2 MV x-ray supervoltage beams, the special "high energy" Victoreen chamber is recommended since it is specifically designed for measurement of this type of radiation.

"Stem errors" may arise with cobalt-60 and supervoltage x-ray beams. The calibration of the Victoreen chamber provided by the

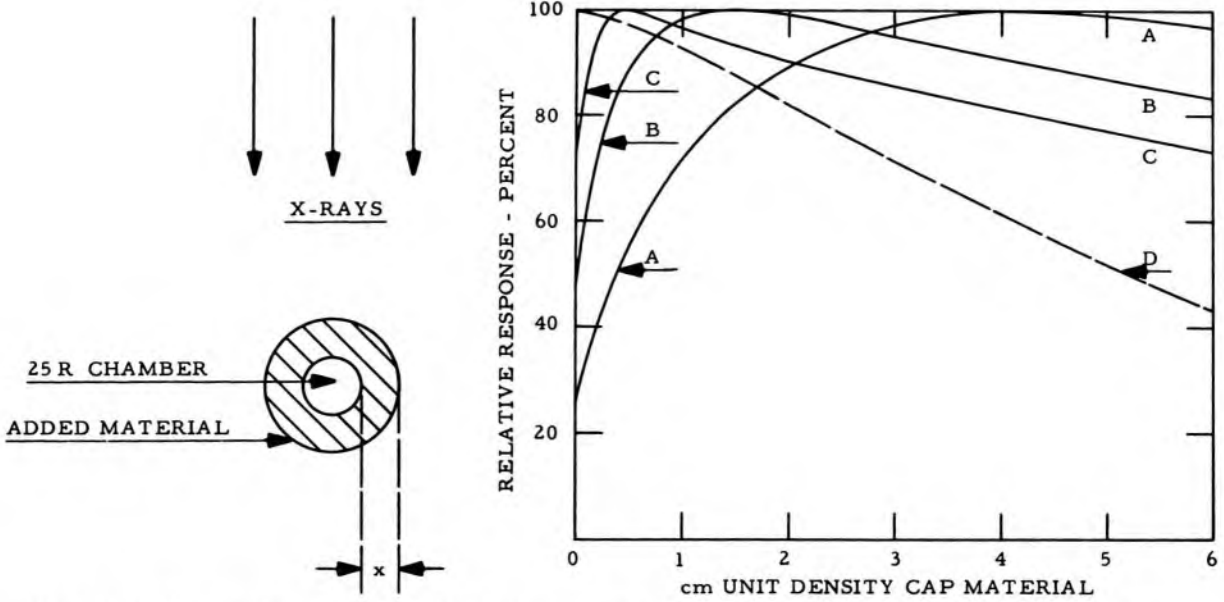


Fig. 11. Typical response of Victoreen 25 R thimble chamber with increasing thicknesses of unit-density added cap material. Left. Experimental setup; x-rays strike chamber surrounded by "x" cm thickness of unit density material. Right. Resulting response, plotted vs. thickness of added cap material, for following four beams: A—22 MV betatron; B—6 MV linear accelerator; C—cobalt-60 teletherapy; D—orthovoltage beam (dashed curve). The orthovoltage curve simply falls with added cap material, from beam attenuation by absorption. With all three supervoltage beams, response first rises to a maximum, then falls due to beam absorption. The required absorber thickness for maximum response increases with photon energy. The maximum represents transient electron equilibrium.

National Bureau of Standards is obtained with the ionization chamber completely immersed in the radiation field. However, in ordinary work smaller fields are used, so the rear part of the condenser chamber unit is rarely, if ever, irradiated. As a result, the N. B. S. calibration is inappropriate without further correction. This is normally taken care of by the physicist in carrying out the supervoltage or cobalt-60 calibration. He must evaluate the stem correction of the unit employed as it varies from chamber to chamber. Errors of several percent in exposure may result if this stem correction is ignored or improperly evaluated.

From the above discussion it is evident that the simple determination of exposure rate from an x-ray generator involves a surprisingly sophisticated process. Although the principles are not inherently difficult, accurate determination of exposure requires the investment of considerable care, time, and money to assure that accuracy requirements are satisfied.

Monitor Instruments

Monitor instruments are usually rate meters used to verify that output from x-ray equipment is relatively constant.¹⁴ X-ray therapy machines should normally be calibrated at least twice a year and additionally checked much more often for consistency of output. Supervoltage machines should be checked daily if possible. Experience shows that, particularly near the end of the life of an orthovoltage x-ray tube, the output may drop fairly rapidly by as much as 30 per cent. This can result from accelerated roughening or pitting of the target, deposits of tungsten on tube windows, etc., as described previously in Chapter 4. In supervoltage machines the problem is even more serious because operating factors may change the output drastically even during a treatment. In such units an instrument is required which is always in the beam, to indicate the total dose received during each treatment. Cobalt-60 or cesium-137 teletherapy machines normally do not require monitor chambers. Except for predictable gradual decay, their output is substantially constant and predictable because they are not x-ray, but rather long-lived radioactive source units.

DESCRIPTION AND USE. Figure 12 shows the location of the monitor chambers in orthovoltage (Top) and 6 MeV x-ray (Right) machines.¹⁵ Note that in both cases they are near the tube, just beyond the filter. [The monitor chamber at top is beyond filter (D); that in the 6 MeV unit is beyond the beam flattening filter (F).]

Older monitor chambers for orthovoltage machines may employ electrodes of high atomic number constituents. Such materials contribute substantial numbers of photoelectrons; these produce extra ionization within the monitor chamber, so the measured current is much greater than otherwise. This permits use of a sensitive galvanometer on the control panel, without any amplifier. Such simple systems are still in use in many older machines and have the advantage of minimum maintenance.

This type of monitor system unfortunately requires a rather large ionization chamber to produce adequate current to operate a galvanometer directly. Modern orthovoltage machines do not have the space in the beam for convenient insertion of such units. Both these and higher energy machines now employ smaller, essentially "air-wall" type ionization chambers (that is to say, with low atomic number electrodes and chamber walls). They produce relatively low current for a given x-ray machine output, and an amplifier must be used in conjunction with the meter. Such instruments, of course, require maintenance of the amplifiers; in some locations this has resulted in serious maintenance problems in the past. In more recent years, how-

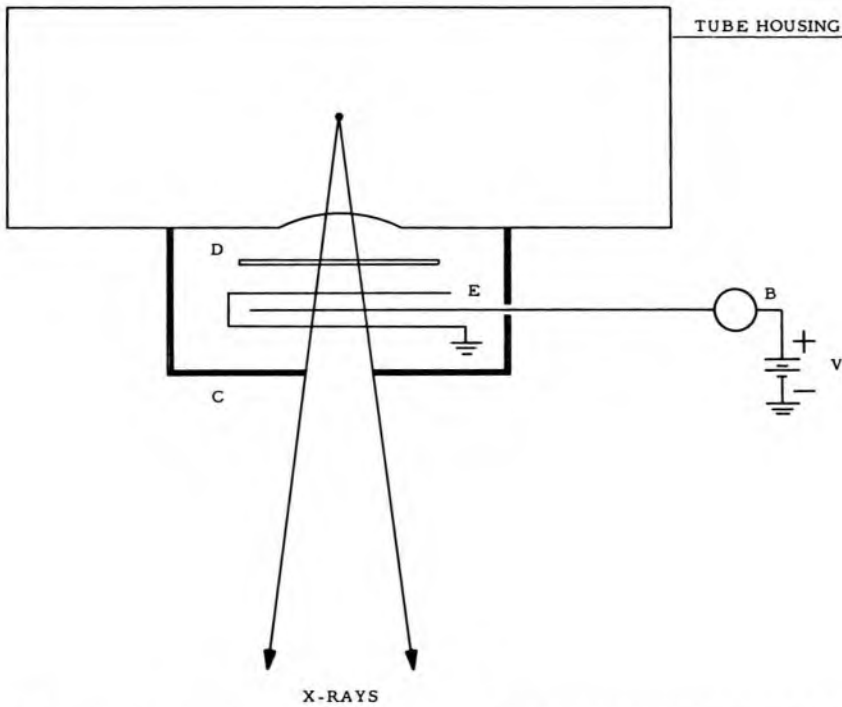
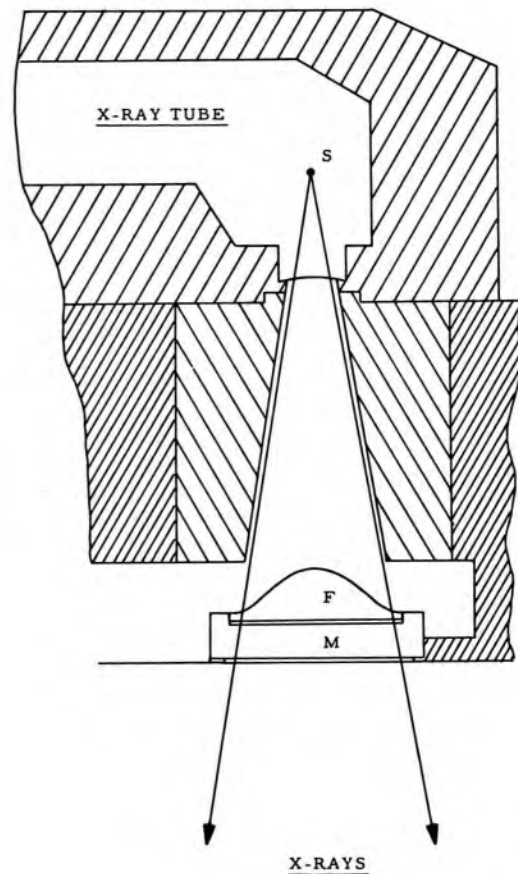


Fig. 12. Monitor chambers: location in orthovoltage and supervoltage machines. Top. Orthovoltage: B—monitor current meter; V—monitor voltage supply; C—collimator (details not shown); D—filter; E—monitor chamber. Right. 6 MV supervoltage: S—source of x-rays; F—beam compensating filter; M—monitor chamber. Note that both monitor chambers are mounted close to irradiated filters. In addition, the orthovoltage monitor is close behind the lead collimator assembly. The chambers, therefore, respond to considerable radiation which never reaches the patient. (Redrawn from Haimson and Karzmark.¹⁵)



ever, most x-ray service organizations offer better service of electronic circuitry, so this problem is less serious than formerly.

Another monitory system requiring no amplifier has been described. It employs a capacitor and pocket electrometer combination, which is an integrating rather than rate meter type instrument.¹⁴

Monitor chambers are preferably of parallel-plate construction in which the x-ray beam strikes the plates perpendicularly to assure uniform beam filtration. Thimble chambers may cast objectionable nonuniform shadows.

LIMITATIONS. Monitor chambers require the same corrections as any other cavity chamber as well as some of their own. Unless they are sealed correction must be made for pressure and temperature. The housing temperature is usually higher than that of the room as a whole, so that the temperature correction may be much greater than in an ordinary application and in addition varies during tube housing warmup. Another effect of the higher housing temperature is that the volume of the chamber itself may change, from buckling of plates, etc. As in any other type of ionization chamber, the materials employed introduce their own energy dependence effect. In monitor chambers this is aggravated by the fact that the x-ray beam impinging upon the chamber has just traversed filtering material. As a result, the chamber is struck by scattered and characteristic radiation from filters which does not reach the patient.

The ionization chamber cannot read as accurately in the monitor location as otherwise for yet another reason: there is back-scatter from objects beyond the monitor chamber. In an orthovoltage machine a monitor chamber reading may increase several percent when the cone is changed from a large to a small size! This is because the lead diaphragm directly beyond the monitor chamber changes with field size, thereby varying back-scatter. Of course all of this scatter is not present at the surface of the patient to the same degree, so the patient dose does not accurately follow changes in the monitor chamber dose.

It must therefore be stressed that *any* monitor chamber cannot possibly yield the accuracy of a careful exposure dose calibration with an appropriate instrument. However, monitor chambers can be quite useful, and in many cases invaluable, to indicate approximate dosage values. If care is taken to correlate actual exposure under treatment conditions with the corresponding monitor chamber dosage, the monitor chamber reading can be quite accurate.

In orthovoltage work monitor chambers cannot normally give a useful indication of patient dosage because of the tremendous energy dependence of the chamber indication. However, they are invaluable in indicating that something is grossly wrong. If readings are recorded

twice daily as a matter of routine, any trends in operating output can be noted early in a course of treatments. Furthermore, incorrect filters can be detected promptly by glancing at the monitor indication before treatment is actually carried out. The writer has on occasion detected not only incorrect but also damaged filters in this way.

DEPTH DOSE MEASUREMENT

Depth dose data are required to estimate ionization distribution within the patient. They are usually measured in rectangularly shaped arrangements of tissue equivalent materials called phantoms. Most often only soft tissue is simulated. Two types, small cavity chambers and extrapolation chambers, are used by physicists making these basic measurements.

In this section we discuss the design and use of phantoms, small ion chambers, and extrapolation chambers for depth dose measurement.

Phantoms

In principle one could perform in-vivo measurements in patients receiving therapy to evaluate absorbed dose. However, such measurements are ordinarily limited to accessible cavities and, in addition, require rather rugged and tiny instruments used under relatively sterile conditions. In-vivo measurements are therefore restricted to special problems.

An alternative is to use cadavers. Although patient hazard is thereby eliminated as a problem, the measured data are not very helpful for several reasons. First, measurements are still normally restricted to accessible cavities. Second, cadavers deteriorate rapidly unless preserved, particularly lung tissue, and the x-ray attenuation characteristics of the body change even after embalming. Finally, x-ray attenuation varies among cadavers of various size and body composition just as among patients, so the data obtained may still require corrections before use, just as those from homogeneous phantoms.

Consequently, homogeneous soft tissue phantoms are generally used. Their use provides standardized depth dose data (isodose charts) of minimum bulk, which can be conveniently stored and consulted when needed for use in treatment planning. Appropriate corrections can then be made as required to allow for tissue inhomogeneities in the living patient, arising from air in lungs, skeletal bone, etc.

Several phantom materials have been employed for making these standardized depth dose measurements. These include paraffin mixtures, sugar plus magnesium oxide, Pressdwood, rice and flour, and water. At present three types of phantoms are most commonly employed in measurement of depth-r. The first is large tanks of water of adequate depth and cross-sectional area. These must be used with probes specially protected to keep them dry. Second, Masonite or Pressdwood phantoms have been used with x-ray films to measure dosage distribution from high energy photon beams. Such phantoms are quite convenient because x-ray film in appropriate envelopes can be inserted between Masonite layers. Unfortunately, there can be significant energy dependence errors when this method is employed, even at relatively high photon energy when one might offhand expect them to be less serious. Finally, a commercial plastic called "mix-D" is often employed for orthovoltage and low photon energy depth dose measurements.

In all phantom materials two physical requirements exist. The mass attenuation coefficient values must vary with photon energy in a manner similar to that of soft tissue, for the relevant energy range; and the density must be reasonably close to that of water.

Of all these phantom materials, water has been most extensively used for orthovoltage and supervoltage beam depth dose measurement while "mix-D" has been employed in the low energy superficial therapy range.¹⁶

Tiny Ionization Chambers

Depth dose data are most often obtained using small ionization chamber probes. These are usually moved across the field of x-rays at a fixed depth within a water phantom. Water is more convenient for this type of measurement than solid phantom materials. In addition, water closely simulates tissue absorption in most clinical x-ray work. The ionization chamber may be moved automatically as well as manually, and automatic isodose plotters have been constructed which greatly facilitate depth dose measurements.

BASIC DESIGN PROBLEMS. Ionization chamber design must take the following into account: chamber size, chamber materials, and perturbation of the x-ray field by any cable used.

Chambers should be fairly small, or they read erroneously. The general specification is that their inside diameter be 5 mm or less with a length of about 1.5 cm or less. Actually, as small as 3 mm in diameter has been employed, but it becomes increasingly difficult to make chambers any smaller than this because serious insulating and fabri-

cation problems arise. In addition, any slight cable motion introduces appreciable dielectric stresses which can produce disturbingly large electrical transient voltages. Both this and stem leakage problems are more troublesome with low than high ionization currents.

From the above, 5 mm appears to be a practical minimum ionization chamber size. An important additional requirement is that the ionization chamber measure rapid intensity falloff at the edge of some x-ray beams. It has been shown that the use of a 5 mm diameter ionization chamber usually introduces negligible error in this measurement.¹⁷

However, each reading still suffers from an uncertainty in the meaning of "depth." Figure 13 shows several ionization chambers whose air volumes are centered at the same depth "d." Note the x-rays in general traverse less than d thickness of phantom or chamber wall material; the rest is air. This error can be minimized by using very small chambers such as (1) and (2), but experimental problems of the type previously discussed may result.

In practice a compromise is made. First, moderately small cylindrical chambers are used to determine the dosage across the field

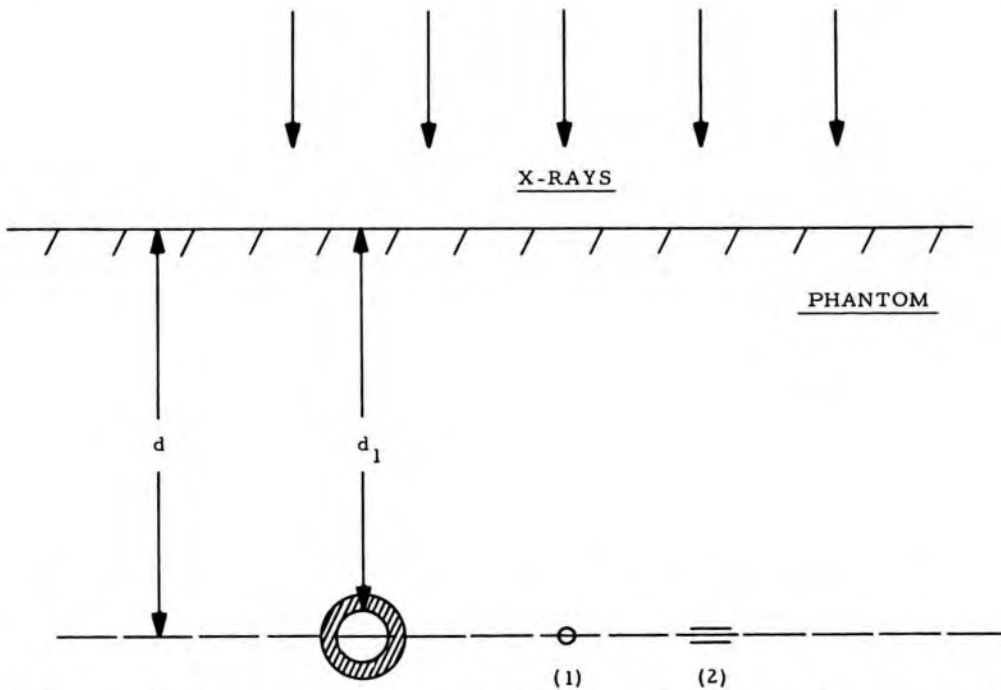


Fig. 13. Error in depth dose measurement with thimble chambers. Chamber at left is centered at desired depth d . Note the x-rays generally traverse less than " d " thickness of material (as little as d_1). The very small chambers at right can minimize the absorption error (1 and 2), but the resulting low chamber current is so small that experimental errors result. The extrapolation chamber was developed to solve this dilemma.

relative to that at the central ray for a given depth. Then the actual depth dose values are computed from more accurate measurements of the central ray depth dose. These are determined by use of an "extrapolation chamber," described below, which is designed to solve the dilemma indicated in Figure 13.

From our previous discussion it is evident that ionization chamber materials must be carefully chosen. The chamber response should be suitable for the full range of x-ray quality found within the phantom. Usually the chamber response must be accurate down into the orthovoltage region even in supervoltage depth dose measurement. This requirement results from the fact that multiple scatter within the phantom greatly softens the x-ray beam at depth, so some relatively low energy photons are measured by the ionization chamber even with supervoltage beams. If the materials of the chamber are not reasonably "air-wall," it responds erroneously. Recall that two requirements exist for the accurate measurement of roentgen exposure in a phantom:

1. The wall thickness must be adequate to absorb all primary electrons produced in the phantom.
2. The ionization chamber must respond accurately *in air* to x-rays of the same quality as are present in the phantom.

The second requirement is usually met primarily by selection of proper materials for the ionization chamber. The first requirement is not normally difficult to meet at lower photon energies when primary electrons do not penetrate materials very greatly. In supervoltage measurements the electrons travel much farther in both tissue and chamber wall materials, and a suitably designed ionization chamber is required to prevent their entering the chamber volume. Under these circumstances, one generally measures a hybrid quantity somewhere between rads and roentgens, and proper corrections should be made to obtain either absorbed dose in rads or exposure in roentgens to avoid confusion. Fortunately the ratio of air and tissue energy transfer coefficients does not vary greatly from 0.1 MeV up to well above 3 MeV, so errors in percent depth dose from ignoring this effect in supervoltage measurements contribute a negligible error clinically.

If one desires *absorbed dose*, the chamber wall should be adequately thick to absorb all primary electrons impinging on it and an appropriate conversion factor employed to convert from roentgens to rads (more accurately, from chamber wall material to tissue).

Any cable with which the ionization chamber is associated is bound to attenuate x-rays differently from the phantom material. The error introduced by this effect can be made negligible by the use of thin coaxial cables now available.

FAILLA EXTRAPOLATION CHAMBER

To solve the dilemma of the thimble chamber in depth dose measurements, Failla developed an extrapolation chamber.¹⁸ Although originally employed for central ray depth dose and back-scatter measurements, the very useful principle has since found many other radiation dosimetry applications. It solves two problems inherent in the use of any thimble chamber. As explained above, there is no exact thimble chamber "position" within the phantom; and a thimble chamber has directional dependence, arising from its cylindrical shape and high Z materials of the associated cable. In general it reads accurately only when the x-ray beam is directed perpendicularly to its central collector.

DESCRIPTION. The Failla extrapolation chamber eliminates these problems by using parallel plates only and avoiding higher Z materials. Figure 14 shows the essentials of such a system. The phantom employed in extrapolation systems is usually a solid one and would normally be of slabs of something like "mix-D" in modern work. The top portion is located at the appropriate treatment distance and made as thick as required to correspond to the desired depth. The lower portion is arranged so that it can be precisely positioned. Its top surface is parallel to the upper portion at a distance "x" which is accurately adjustable to as small as a few millimeters. The lower surface of the upper portion is coated with a thin layer of colloidal carbon to render it conductive, as are parts of the upper surface of the lower portion. The latter surface is divided into two basic areas: a central, round collecting electrode C, surrounded by the rest of the surface G, which is grounded and insulated from C. A collecting voltage V is impressed across B and C and a null type detector used to measure the collected charge (similar to that in the free-air chamber).

Note that the actual volume in which the collected ionization is produced is a right cylinder of volume (Ax), where A is the effective area of the collector electrode C. Ionization occurring between B and G is of course collected without being measured. Precautions are taken to assure voltage saturation.

PROCEDURE. The procedure in use of this instrument is as follows. The gap "x" is adjusted to a desired value and the voltage V properly adjusted. The machine is turned on for an appropriate time and the charge collected. The charge is then divided by the total volume Ax and the result plotted versus x (Fig. 15). The procedure is then

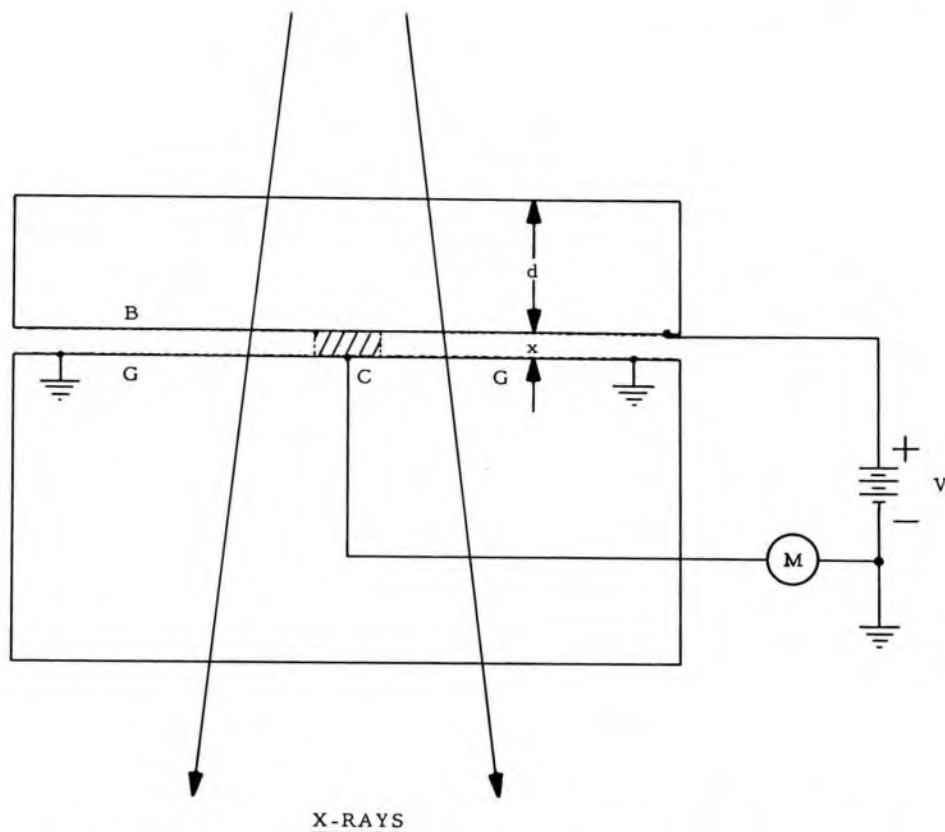


Fig. 14. Failla extrapolation chamber (see text). Beam traverses thickness d of phantom, an air gap, then the rest of the phantom. C is central collector, with surrounding guard plate G. ("Plates" are thin graphite coatings on phantom materials.) Collecting volume (shaded volume) is Ax ; A is the effective collector area, and x is separation of upper and lower parts of phantom. Adjustments are provided, to keep plates parallel and to permit varying x at will. Charge measurement: V is the collecting voltage, enough to assure saturation voltage, and M is the null detector system to measure collected charge.

repeated for several other plate separations both smaller and larger, so that one can draw a graph joining these points. Finally, the curve is extrapolated to zero plate separation. The esu/ml thus determined from the graph is the actual measured roentgen value (37.7 R in the figure). This need be corrected only for pressure and temperature to give the true roentgen dose, and the rad dose is then easily determined by use of the appropriate phantom material conversion factor.

If one tries to take short cuts and simply utilizes the esu/ml ratio at 5 mm separation, an erroneously *low* value of roentgen exposure is obtained because all of the air in the volume is below the desired depth, as is the lower part of the phantom. As a result, the lower part of the air volume receives a lower *incident* x-ray intensity and consequently has less ionization than the upper part of this volume. Also, the lower part of the phantom, being farther away from

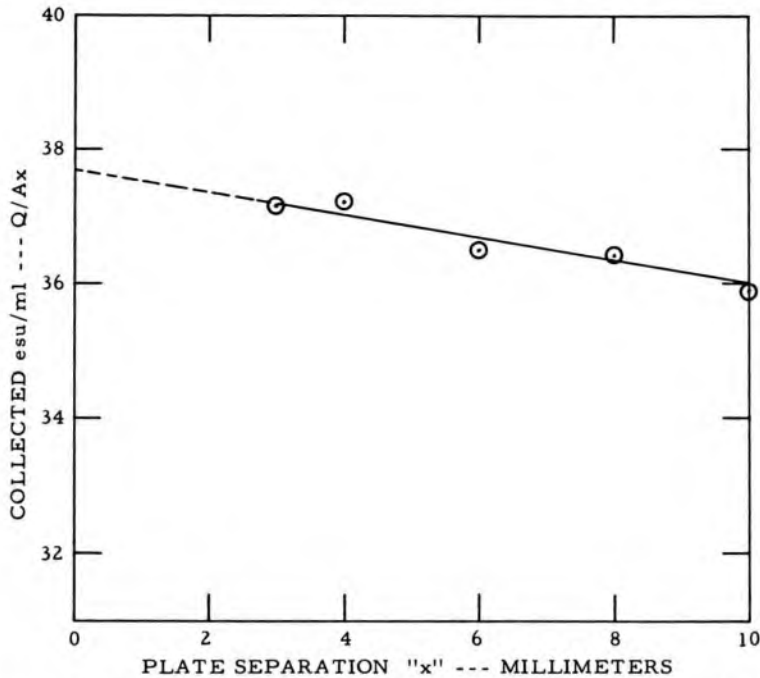


Fig. 15. Curve illustrating extrapolation procedure. Circled point data represent measured esu/ml. [Q is number of esu produced in shaded volume of Figure 14 (Ax)]. A nearly straight line curve is drawn through these points and extrapolated to zero separation, yielding 37.7 measured R in the example shown. Note special vertical scale to increase plotting accuracy.

the source than when it is in contact with the upper portion, receives x-rays of lower intensity and thereby contributes less x-ray scatter *back* into the volume. As a result of these two effects, the measured esu/ml ratio *increases* as the plate separation is *reduced*.

One might ask, "Why not simply use a flat chamber which has extremely tiny separations of the two electrodes?" As indicated previously, such a chamber would be prone to all kinds of electrical instability problems; in addition, its volume would be hard to measure accurately. By the extrapolation procedure one can make very accurate readings and extrapolate them back into the experimentally troublesome region. This is the beauty of any type of extrapolation procedure—it permits obtaining a difficult experimental result by combining relatively simple measurements and graphical procedures.

OTHER RADIATION DETECTORS

Our discussion so far has referred only to ionization chambers, although many detectors employing other radiation effects have specific uses in radiology. These are briefly summarized below with comments as to their nature and applications. (See Table 2.)

TABLE 2. OTHER RADIATION DETECTORS

General Method	Device	Application
A. Chemical		
1. Solutions	1. Fricke dosimeter— $\text{Fe}^{++} \rightarrow \text{Fe}^{+++}$	1. Standardization of energetic x-ray and electron beams. Useful range 5,000–50,000 rads.
2. Photographic film	2. Film, exposed in darkness	2. X-rays: good for air-R, but keV effects in phantoms Electron beams—very useful.
B. Luminescence		
1. Fluorescence	1. a. Probe scintillation detector (ZnS, anthracene, etc.) b. Scintillation detector, NaI (Tl)	1. a. Sensitive but keV dependent. Varies with both crystal material and size. b. Basic to modern radioactivity work
2. Delayed; heat-released	2. Thermoluminescence (TLD). LiF has best energy response	2. Very promising. Reproducibility a potential error source.
3. Delayed; released by ultraviolet light	3. Photoluminescent glasses (silver-activated phosphate glass)	3. Energy dependent, used for radium dosimetry mainly.
C. Electrical		
1. Direct conduction	1. CdS crystal	1. Sensitive, needs simple circuitry. Problems: crystals very variable, of sluggish response, energy dependent, limited accuracy.
2. p-n and p-i-n junction units	2. Lithium drift and other silicon diodes	2. So far, mainly particle detectors. Recent promising developments.
D. Calorimetry	Special laboratory units.	Basic standardization of high energy x-ray and electron beams.

Chemical Detectors

Ionization and excitation following x-ray absorption produce chemical changes which can be used for dosimetry. We shall consider effects in inorganic aqueous solutions and photographic films.

CHEMICAL SOLUTIONS. The biologic response which one would like to infer from the absorbed dose is basically a chemical effect, so it would appear desirable to use a chemical method for measuring dosage. One of the most useful methods developed is the so-called Fricke dosimeter which utilizes the oxidation of F^{++} to Fe^{+++} in a dilute sulfuric acid solution. Unfortunately, the required dosage levels are quite high for any degree of accuracy. The system, even with its more sensitive modifications, will not respond well to doses much below 10,000 rads. In addition the method is sensitive to oxygen concentration. Another system that has been used employs the reduction of ceric sulfate to cerous sulfate. This method is less sensitive than the Fricke procedure but relatively independent of changes in oxygen concentration. Both techniques are very critical as to their execution and require well-standardized chemical procedures.

Other chemical methods have been developed which are more sensitive but correspondingly more troublesome to carry out. Currently chemical methods are of interest primarily to the researcher, with the exception of the Fricke method which is employed for absorbed dose standardization in the 6 to 42 MV x-ray and electron beam region.

PHOTOGRAPHIC FILMS. Photographic film is one of the oldest radiation detectors, having been used diagnostically since the discovery of x-rays. It contains silver bromide, and both silver and bromine have relatively high atomic numbers. Consequently, the radiation sensitivity of x-ray film varies considerably with photon energy. For example, it takes about 30 times more rads from 1 MeV than from 40 keV photons to produce the same film darkening! Great care must be taken, therefore, in interpreting density distributions on films exposed to x-rays, particularly in the orthovoltage range.

X-ray films may be usefully employed to measure supervoltage x-rays, but one must be quite careful even in this application.²⁰ For example, consider Figure 16(Side) which indicates an x-ray film sandwiched between two portions of a phantom and exposed edgewise. This appears to be a convenient way of determining a great deal of information with a single exposure. Figure 16(Below) relates the actual x-ray distribution at depth to the indicated distribution determined on the basis of film density for a particular 10×10 cobalt-60 beam at 70 cm treatment distance. The bottom curve (A) shows the distribution of *directly transmitted* radiation dose at this depth across the field. The solid curve (B) gives the total gamma-ray intensity at the film location; this includes both the directly transmitted and scattered radiation.

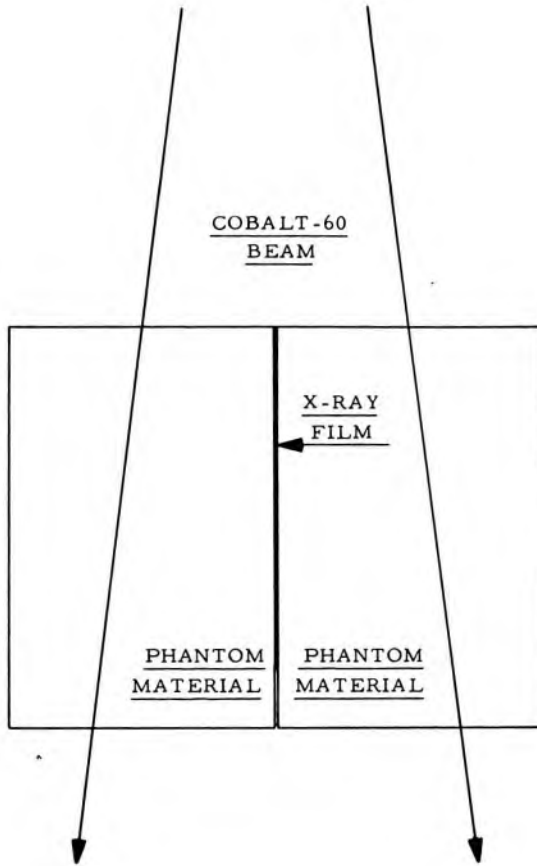
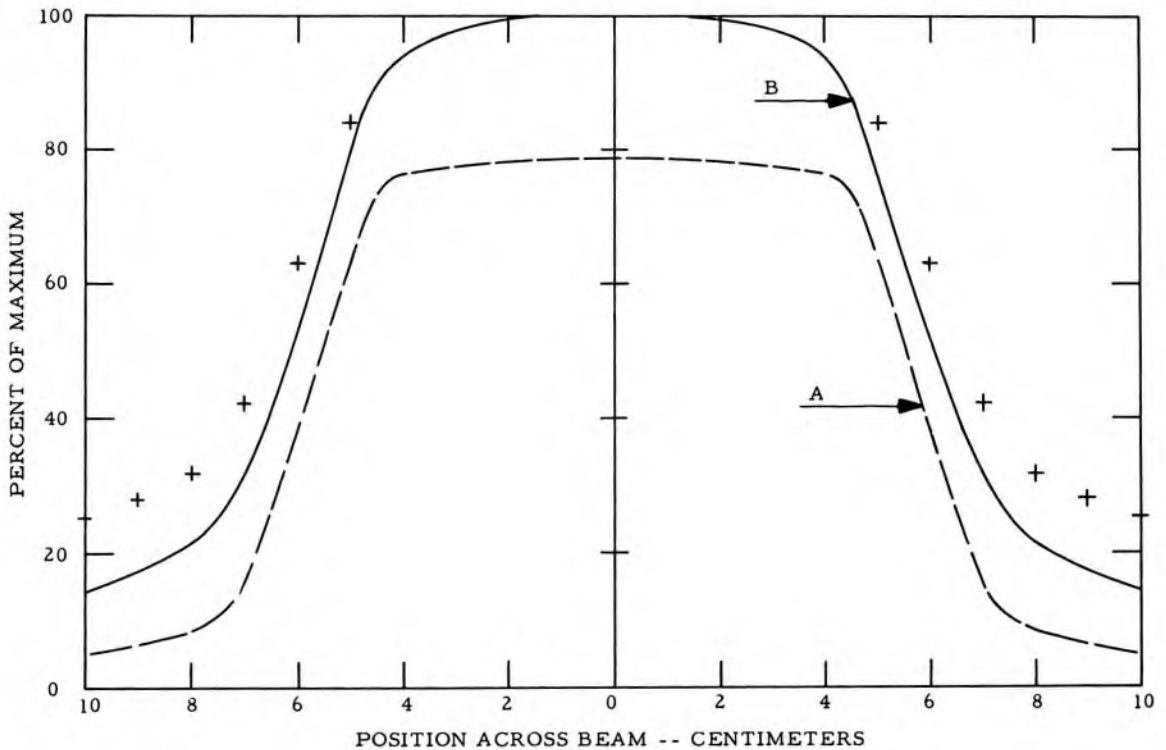


Fig. 16. Supervoltage depth dose measurement using film. Left. A popular technique of exposure. Film is sandwiched between halves of the Masonite phantom, exposed edgewise. Below. Data across field at 10 cm (10 x 10 cm field, 70 cm SSD, cobalt-60 beam). Lower curve: Beam directly transmitted to 10 cm depth; contributory scatter not included (computed). Solid curve: Actual measured dosage distribution across field; includes both directly transmitted and scattered radiation. Cross data: Indicated dosage distribution across field, using film (corrected for its nonlinear response). The film response is disproportionately high at the edges of the field, indicating a falsely large field width. (Redrawn from Stanton.²⁰)



There is a significant increase in dose due to contributory scatter in the *center* of the beam, primarily because of forward and side scatter. At the side there is considerable scatter present in the marginal regions of the beam, probably due to side scatter. The crosses above the solid curve show the dosage distribution across the field indicated by film density readings. Notice that a substantial error in measured field size results because the film responds disproportionately to radiation near the *margin* of the field, which is softer on the average than the radiation in the *center* of the field. This quality difference reflects the fact that radiation at the edge consists of almost all scatter, and scattered radiation is softer than direct. (In the center of the field, as indicated by the two lower curves, scatter constitutes only about 20 percent of the total intensity whereas it constitutes closer to 80 to 100 percent of the intensity near the margin of the beam.)

It is evident, therefore, that one should use film in phantoms with cautions to measure x-rays. A technique has been developed for obtaining accurate single-field isodose distributions in phantoms, but significant corrections are required to achieve accuracy approximating that of ionization chamber measurements. In general, scatter errors are proportional to the beam penumbra.²¹

Film yields quite accurate results in measurements of distribution of x-ray exposure in air. Because scatter effects are then minimal, all parts of the film are exposed to x-rays of essentially the same quality, so the response is reasonably the same over the entire film. Also, film dosimetry of high energy electron beams yields accurate results even in phantoms.

In any film dosimetry, film development must be carried out with great care, otherwise errors can result in the measured distribution.

Luminescence Detectors

Many materials produce light (are luminescent) because of fluorescent phenomena. Besides ordinary fluorescent effects there are others in which crystals are excited by some event like a photon or electron interaction but cannot release their characteristic light photon without help. This help can be given later in the form of heat (thermoluminescence) or ultraviolet light (photoluminescence).

FLUORESCENCE. Fluorescent detectors were employed for diagnosis early in x-ray work. As previously indicated fluorescent screens use zinc sulfide, and intensifying screens, cadmium tungstate. Sodium iodide crystals activated with thallium are extensively used to detect gamma rays from radioactive substances (Chap. 10).

Fluorescent probes have also been used for x-ray dosimetry. High atomic number detectors are more sensitive than low but exhibit considerable energy dependence and are therefore poor for most depth dose work. Anthracene and other organic materials have been used with zinc sulfide for orthovoltage x-ray dosimetry. Fluorescent probe dosimeters have not been as reproducible in practice as ionization probes and are consequently not often used nowadays for depth dose measurement.

THERMOLUMINESCENCE. In recent years much work has been done with certain alkali halides which exhibit the phenomenon called "thermoluminescence." These are usually crystals of either magnesium or lithium fluoride, which respond in a very peculiar manner when irradiated by x-rays. Although they appear to be unaffected at the time of irradiation, they store energy. When suitably heated, irradiated crystals release stored energy as light. The total amount of light as measured by a photomultiplier instrument is roughly proportional to the dosage to which the crystals were exposed.

A great deal of work is currently in progress on the use of lithium fluoride, which is relatively energy independent because of the low Z of its components. Reproducibility of measurement of better than ± 5 percent has been reported. Great care must be employed in handling these materials, however, because they also fluoresce when mechanically rubbed or jarred, producing handling errors. A promising development is Teflon-sealed lithium fluoride detectors in which variables of mechanical dispensing and handling are eliminated by plastic mounting of the crystals in discs or 1 mm rods. Thermoluminescent materials also exhibit memory effects, and the thermal cycle is critical.

PHOTOLUMINESCENCE. Finally, there is another category of solid materials called "photoluminescent" detectors. These are a group of silver-activated phosphate glasses, which have been employed to measure radium dosage. When exposed to x- or gamma rays such glass stores some of the energy, just like lithium fluoride. However, instead of using heat to release this energy, one uses ultraviolet light. Visible light is released, the amount depending only on the radiation dose to which the glass has been exposed; it is separated from the incident ultraviolet light using appropriate instruments.

Electrical Detectors

Cadmium sulfide and p-n junction diode crystals have been employed more and more recently for radiation detection.

Cadmium sulfide crystals decrease in electrical resistance when irradiated by x-rays. They are sufficiently sensitive that this

effect can be used directly as a measure of x-ray irradiation without amplifiers—only a battery and meter are required. They have several disadvantages, however, including a marked time delay in response and an inconvenient nonlinear scale of dose versus meter reading. In addition there is a great variation in sensitivity among crystals. However, technologic developments may result in improved units, and selected crystals are currently used for gamma ray measurement and some control applications.

The p-n junction diodes are related to transistors. They are generally made of a single semiconductor crystal, usually silicon, specially processed with various elements to create certain crystal discontinuities. When irradiated these units produce both a current and a voltage akin to that of a photovoltaic cell. Either phenomenon may be measured to obtain absorbed dose, but the current use is more accurate. These units are more energy-dependent than ion chambers because the lowest atomic number material employed is silicon. However, these p-n junction diodes have been usefully employed recently to measure depth dose from supervoltage radiation.

A newer modification is the p-i-n junction diode. These units are far more sensitive than p-n junction units. They have potential application for measurement of low intensity x-rays and light, as well as for high resolution spectrometry of particle and x-ray photon beams.

Calorimetry

Radiation calorimetry measures the total energy in a particle or x-ray beam and is used to standardize high energy x-ray and electron beams. These are allowed to impinge upon a block of some very effective x-ray absorber, such as lead, and the subsequent rise in temperature is noted. Rather elaborate procedures must be employed to measure the very, very slight temperature increases involved, so this is a laboratory procedure. Such measurements are primarily for the purpose of determining fundamental dosage characteristics of ionizing radiation, particularly in the energy range where ionization chambers have some practical limitations: below 5 keV and above 2 MeV.

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7

Clinical X-ray Dosage Evaluation

Published depth dose tables and isodose charts are basic tools used to evaluate patient dosage in radiotherapy. This chapter discusses the nature and use of these tables and charts, following a brief section on basic dosage concepts and definitions. As indicated in Chapter 5, integral or volume dose vitally affects the patient's ability to tolerate radiotherapy, so a short section is also presented to show how volume dose may be calculated. Finally, supervoltage x-rays are preferred nowadays to orthovoltage for radiotherapy of almost all treatment sites but the most superficial. The reasons for this preference are examined in the last section, and a brief discussion is included of high energy machines and the beam shaping filters employed in their use.

CONCEPTS AND DEFINITIONS

Radiation therapy planning involves two basic steps. The first is purely medical and starts with a thorough workup of the patient. This includes not only his general medical status but also the character, distribution, and likely mode of spread of his disease. In consultation with medical and surgical colleagues the radiotherapist makes a decision as to the best approach for the particular patient: irradiation, surgery, chemotherapy, or combinations of these. To an increasing extent the decision involves definitive radiotherapy, a procedure just as serious to the patient as definitive surgery, and demanding considerable skill and care in its planning and execution.¹

The second step is preparation of a treatment plan. Central to this is a medical decision. How many rads are to be delivered, where, and in what over-all time? Calculations are often useful to find the optimum combination of x-ray treatment fields and radioactive

sources to achieve a desired dose distribution, and the services of a radiologic physicist are helpful, particularly in uncommon situations.

Basic to all calculations is the relationship of the incident beam intensity to dosage within the patient, as well as dose distribution around any radioactive sources employed. Dosage from radionuclides is covered in Chapter 12. Depth dose tables and isodose charts are used to compute x-ray dosage distribution. Before discussing dosage tables and charts in detail, several dosage terms must first be defined and the relationship between roentgens and rads previously mentioned must be more clearly established.

Clinical Dosimetry Terms

Figure 1 shows an x-ray beam impinging upon the surface of a patient or phantom. Two types of terms are employed to describe the situation, the first relating to a description of the beam itself and the other to dosage within the patient.

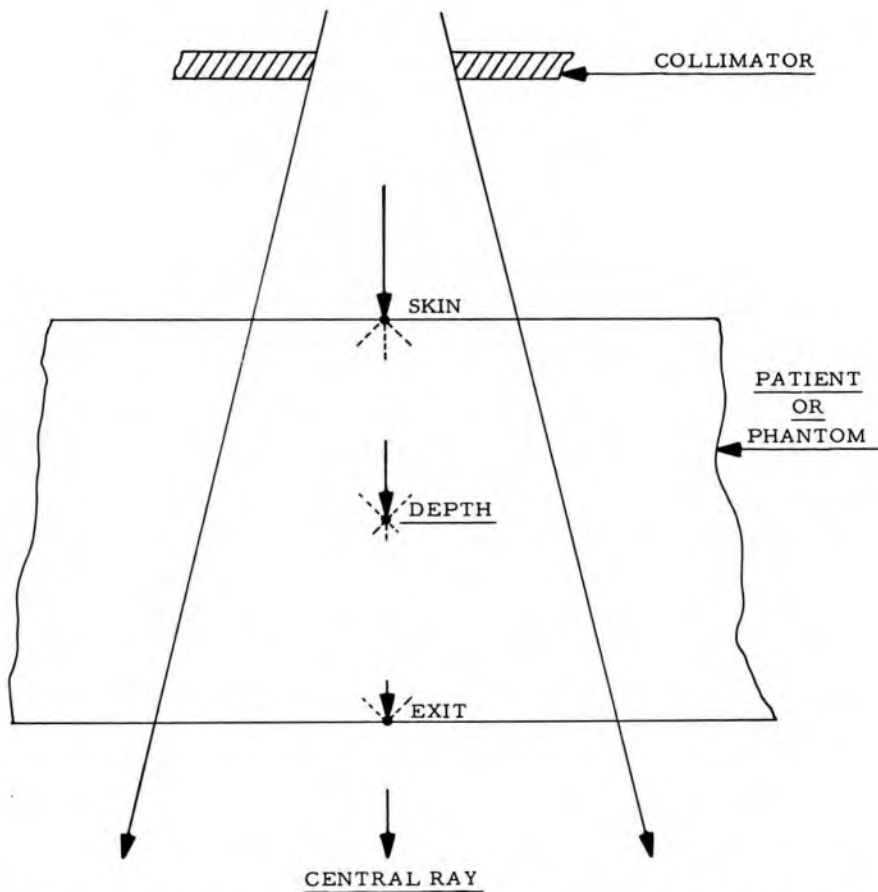


Fig. 1. Definition of central ray dose terms: central ray, air, skin, depth, and exit dose (see text). Solid lines with arrows are direct x-rays from source. Dotted lines without arrows are scattered x-rays.

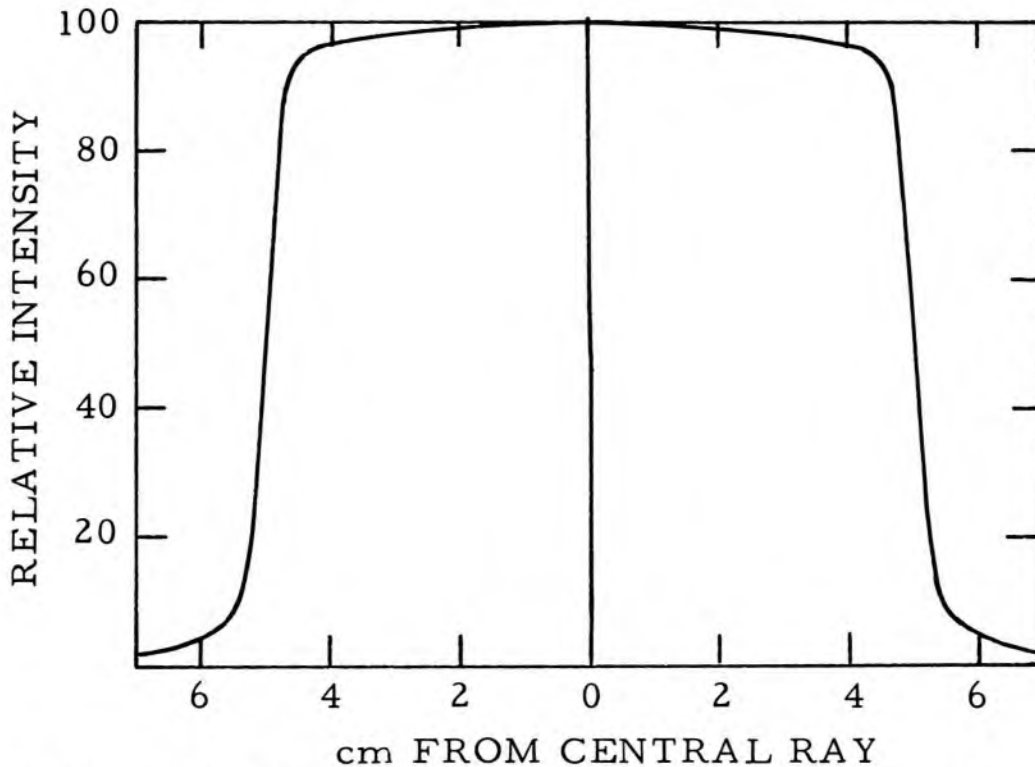


Fig. 2. Dosage distribution in air across a well-collimated x-ray beam.

THE TREATMENT BEAM. Consider terms relating to the beam itself. The *central ray* is a line joining the source and the center of the beam of x-rays and represents the line indicating the direction in which the beam is aimed. The *source-skin distance* (SSD) refers to the distance from the front surface of the source, which is the tube target or the face of a teletherapy source, to the surface of the patient or other object being irradiated.

Some ambiguity exists in identifying the geometric edge or margin of the field because the beam intensity does not fall off abruptly. Figure 2 shows how the intensity of an x-ray beam measured in air varies across the beam, perpendicularly to the central ray. Note that the beam trails off gradually at the sides. This is primarily because scatter reaches the patient from the source itself and any type of field-defining system; also, some scattered and characteristic radiation emerges obliquely from filters and collimators. These both contribute x-rays outside the nominal margin of the field.

In addition, some sources are relatively large. Although the typical radiation source in an x-ray tube used for therapy is of the order of 3 to 8 mm across, radioactive cobalt sources range from 1 cm to more than 1 inch, while cesium-137 sources are even larger! As a result, there is sometimes a very gradual falloff of intensity near the margin of beams from these sources, usually referred to

as a large "penumbra." Consequently, the *margin of the beam* is by convention taken to be the location where the intensity falls to 50 percent of that in the central ray, with the beam measured in air. (The *penumbra* is often taken as the separation of the 20 and 80 percent of central ray intensity locations in air.) We shall have more to say later about the problem of controlling penumbra in cobalt-60 machines.

DOSAGE TERMS. Now let us define terms referring to "dosage" (Fig. 1). *Air-R* refers to the x-ray exposure during treatment, measured in air, delivered at the location where the skin is to be placed. This is an indication of the x-rays *from the machine* which strike the patient, as shown by the solid arrow directed toward the skin. However, during treatment part of this radiation is scattered. Some rays are deflected only slightly but others sufficiently to re-emerge in the direction from which they came. This is called "backscatter." It differs from the incident x-rays primarily in direction and therefore also contributes to ionization of skin tissue. The roentgen dose from backscatter is quite naturally called *backscatter-R*.

It is evident that the skin receives not only the air-R coming from the tube but also the backscatter-R from the patient. As a result, the *skin-R* or *surface-R* is the air-R plus the backscatter-R. This is important because it means skin dose is increased by backscatter over what one might normally expect; the extent of this increase is typically of the order of one-third for orthovoltage but much less (5 percent or less) for supervoltage therapy.

The roentgen dose below the skin is referred to as *depth-R* and that at the exit surface as *exit-R*. All of these terms apply to doses along the central ray except for depth-R which can be anywhere within the patient.

Some *relative* terms are often used to indicate the extent of backscatter and penetration of the x-ray beam. *Percent backscatter* is the backscatter-R produced by 100 air-R incident on the skin. It follows from the definition of the term "percent" which is Latin for per hundred; thus, 30 percent backscatter means 30 backscatter-R *per hundred air-R*. Another term used to indicate the extent to which x-rays are backscattered in a patient is the term "backscatter factor," which is simply the surface-R divided by the air-R. For example, if we have 30 percent backscatter, 100 air-R incident on the skin results in 30 backscatter-R and hence a total of 130 surface-R. The backscatter factor is therefore 130/100 or 1.30.

The term *percent depth dose* is similarly defined. It refers to the effective roentgen penetration of the beam. For example, a certain 200 kV x-ray beam used at 50 cm SSD delivers 36 R at 10 cm depth

along the central ray, when a total of 100 R is delivered to the surface of the patient. We then have delivered 36 R at depth per 100 surface-R, so the percent depth dose is 36 percent. This type of data is given in central ray depth dose tables. More general data relating to other locations within the patient are given in "isodose charts."

In orthovoltage work maximum ionization density occurs at the skin surface. However, in supervoltage and teletherapy applications, one can no longer use the surface-R as an indication of the maximum dose delivered to the patient. As we have already seen in Chapter 5 (Fig. 6), the *rad* dosage in such situations reaches a maximum *below the skin*. For this reason supervoltage depth dose tables and isodose data refer to rad dosage in water or soft tissue only. Data refers to the *maximum absorbed dose* in tissue, which usually occurs several millimeters below the surface for such beams.

Rads versus Roentgens

It is frequently necessary in radiotherapy to compute the absorbed dose in rads from exposure in roentgens at a given point in the patient. In general, this is given by the following relationship:

$$\text{Number of rads} = (f) \times \text{number of roentgens} \quad (7-1)$$

The factor "f" depends on the absorption characteristics of the tissue involved. Generally this involves the ratio of the mass energy transfer coefficients $(\mu/\rho)_{\text{en}}$ of tissue and air:

$$f = 0.869 \frac{(\mu/\rho)_{\text{en}} \text{ of tissue}}{(\mu/\rho)_{\text{en}} \text{ of air}} \quad (7-2)$$

The term "mass energy transfer coefficient" has been previously defined (Chap. 4).

The $(\mu/\rho)_{\text{en}}$ quantities vary with photon energy as well as absorber material, so f also depends on these quantities. In practice the x-ray spectrum is continuous, and tissue composition depends on its type (i.e., bone, fat, etc.) The symbol \bar{f} is therefore used to represent the factor actually appropriate in clinical work.

Equation (7-2) comes from the definitions of the roentgen and rad. These units involve energy transfer to air and tissue, respectively. It is evident from its definition that the greater $(\mu/\rho)_{\text{en}}$, the more ions are produced in a given mass of material. What of the 0.869 term? It is the number of *rads* delivered by one roentgen to *air*. It differs from unity because the rad and roentgen differ in how energy density is measured: (100 ergs/g versus 1 esu/.001293g).

TABLE 1. FACTORS FOR CONVERSION FROM ROENTGENS TO RADS (\bar{f}) FOR THREE ABSORBERS*

Photon Energy (MeV)	Water	Compact Bone	Muscle
0.010	0.912	3.55	0.925
0.015	0.889	3.96	0.917
0.02	0.879	4.23	0.917
0.03	0.869	4.39	0.911
0.04	0.879	4.14	0.920
0.05	0.892	3.58	0.926
0.06	0.905	2.91	0.929
0.08	0.932	1.91	0.940
0.10	0.949	1.46	0.949
0.15	0.962	1.05	0.956
0.20	0.973	0.979	0.963
0.40	0.966	0.928	0.955
0.60	0.966	0.925	0.957
0.80	0.965	0.921	0.957
1.0	0.965	0.919	0.957
1.5	0.964	0.921	0.957
2.0	0.965	0.921	0.955
3.0	0.962	0.929	0.955

*Data from Table 1, N.B.S. H. 87,² p. 10.

SOFT TISSUE. Table 1 shows \bar{f} values for water, compact bone, and muscle at several photon energies.² Note that \bar{f} is relatively constant for both water and muscle over a wide energy range. Evidently one can ascribe an average value \bar{f} of $0.94 \pm .03$ to muscle for the *entire* clinically relevant photon energy range, so a simple and relatively accurate numerical conversion is possible from roentgens to rads for clinical work. For supervoltage radiation the \bar{f} value is quite constant (about 0.95 through 0.96 for 100 to 2,000 keV); in well-filtered orthovoltage beams the variation is a bit more (0.92 through 0.96 from 40 to 200 keV).

BONE. Bone is quite different. The value \bar{f} varies greatly with photon energy for compact bone because it contains calcium and phosphorus which have high photoelectric absorption coefficients. Consequently, the *rad* dose within bone is much higher than that to surrounding soft tissues—even when the *roentgen* dose is the same! For example, consider a thin flat bone irradiated by 60 keV photons [Fig. 3(Top)]. Of interest are rad doses to four locations: in adjacent soft tissue (A), in compact bone (B), and in the edge and center of the marrow cavity (C) and (D). Assume the bone is thin enough

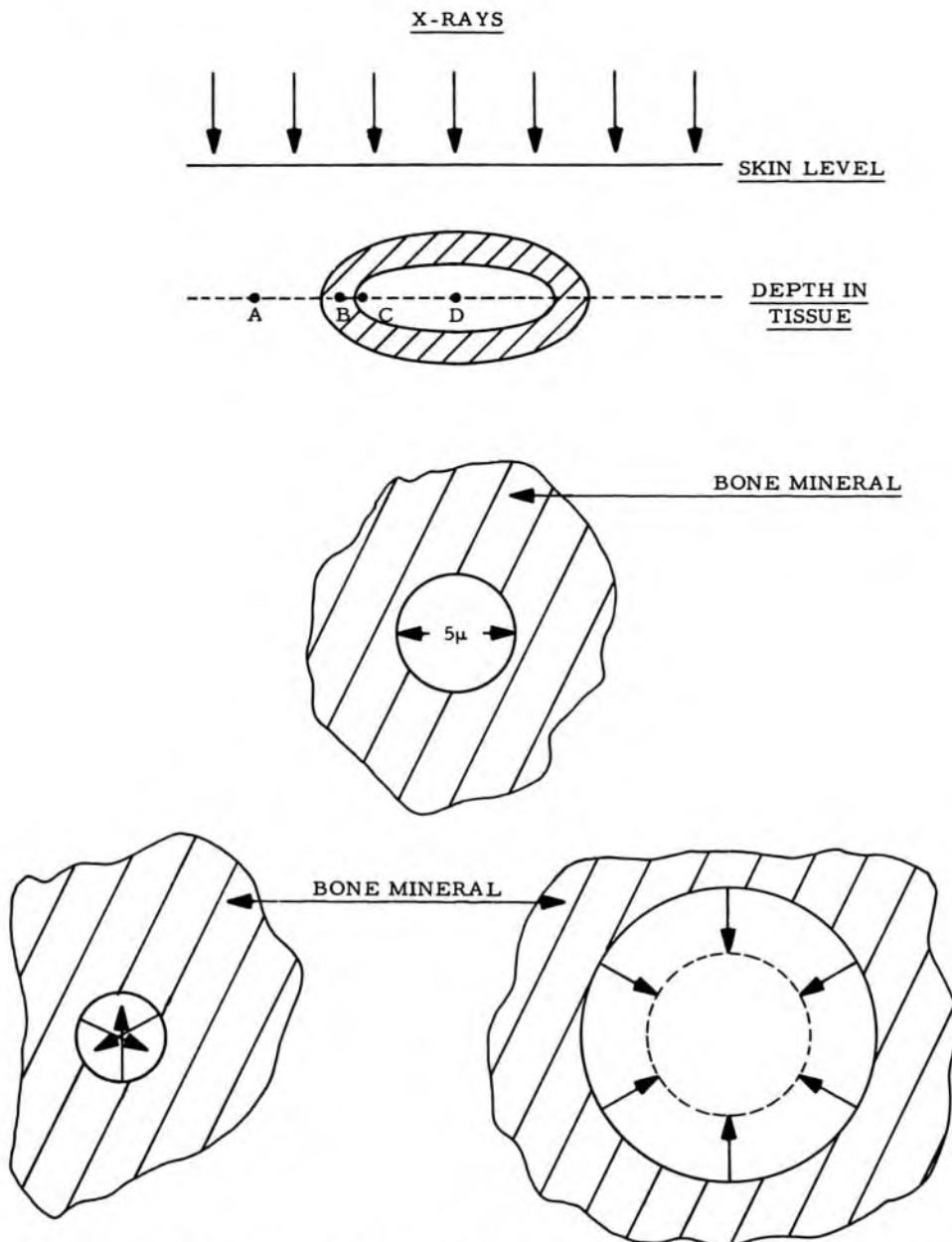


Fig. 3. Absorbed dose in bone³ (see text). Top. Dense bone in tissue—relevant cell locations: A, outside bone; B, in dense bone; C, in marrow space, near bone mineral; D, in marrow space, away from bone mineral. Center. Single osteocyte substantially surrounded by bone mineral. Bottom. Influence of Haversian canal diameter on dose to cells. Left, small diameter vs. photoelectron range. Right, large diameter vs. photoelectron range.

so x-ray doses in *roentgens* are the same at all four locations. (There is, for example, only 1 to 2 percent roentgen reduction when 60 keV x-rays traverse 1 mm of dense bone.) Thus, any difference in local ionization is primarily due to \bar{f} differences.

The rad dose at A is simply 0.93 times the roentgen dose, from (7-1), but that at B is 2.91 times this great! But how can this be? Figure 3(Center) illustrates the answer. It is a greatly enlarged

sketch of a lone osteocyte (roughly 5μ in diameter) surrounded by bone mineral. Were it surrounded by soft tissue it would be ionized primarily by Compton recoils, and 1 roentgen would deliver 0.93 rads. Substitution of bone mineral, however, not only increases the number of recoils (bone mineral is denser) but *also* adds photoelectrons in substantial amounts. The osteocyte, therefore, receives 2.91 rads from a roentgen, over 3 times higher than the rad dose it would receive if surrounded by soft tissue.

More generally, even dense bone has Haversian systems with vessels of varying diameter filled with circulating blood. The latter is transient, so its cells are spared most of the irradiation. Vessel walls, however, receive higher doses, particularly the bone capillaries. Figure 3(Bottom) illustrates why bone capillaries are particularly vulnerable. It shows adjacent small and large vessels exposed to the same x-ray beam. Assume the photoelectrons have a maximum energy equal to the capillary diameter. The small vessel wall is then hit by photoelectrons from *both* the adjacent bone mineral and that on the opposite side! On the other hand, the larger vessel (2) is big enough so photoelectrons give up their energy to the blood before they can reach the other side. As a result, vessel wall rad dosages are lower. It is evident that f in an Haversian vessel varies greatly with the diameter.

Returning to Figure 3(Top), it is evident C receives dosage similar to that at the wall of a large Haversian canal. Because photoelectron ranges are limited relative to the cavity diameter at low energies where the effect is important, most bone marrow cells (D) are not so heavily irradiated, with \bar{f} numbers approximating those of soft tissue.

Table 2 summarizes these results with an estimate for both 50 micron and "average" Haversian canals.³ What practical conclusion can be drawn from all this? One can certainly say that low energy beams deliver disproportionately high rad doses to bone cells, particularly to osteocytes in dense bone and to bone capillary walls. Hence, osteonecrosis is more likely to occur with orthovoltage beams of low than high filtration. Use of low HVL radiation near bone is therefore generally poor practice, and supervoltage is preferable to even well-filtered orthovoltage x-ray beams for such work.

Central Ray Data

As indicated previously, one uses central ray depth dose data to compute dosage to both skin and deeper tissues from the exposure time and beam intensity of each machine. In orthovoltage work the intensity is usually measured as exposure rate at the skin location.

TABLE 2. AVERAGE FACTORS FOR CONVERSION FROM ROENTGENS TO RADS (\bar{F}) FOR SOFT TISSUE COMPONENTS OF BONE ¹³

Photon Energy (keV)	Osteocyte 5 μ Dia.	Estimated "Average" Haversian Canal	10 μ Lining of 50 μ Dia. Haversian Canal
25	2.80	1.73	1.50
35	3.12	2.05	1.76
50	3.25	2.27	1.89
75	2.40	1.85	1.60
100	1.52	1.36	1.26
200	1.05	1.035	1.02

Skin-R is then obtained from air-R using percent backscatter or backscatter factor data, and depth-R using depth dose data. In supervoltage and to an increasing extent in orthovoltage work, the machine output is measured in terms of the *absorbed dose* rate at a suitable depth (5 cm, usually) in a water phantom for the particular exposure conditions and depth dose data used as before.⁴ At this writing the most reliable sources available are given in the British Journal of Radiology Supplement 10.⁵

Backscatter

As with any other kind of scatter, backscatter depends essentially on two quantities: the x-ray quality and amount of scattering material. The latter is determined in radiotherapy by the size of the irradiated field and depth of underlying tissue. From 8 to 20 cm of tissue are required for maximum backscatter to occur, for radiation in the clinical range of qualities.

Table 3 shows how percent backscatter varies with beam field size and HVL. Percent backscatter increases continuously with the field size, for all beam qualities. The increase is greatest, however, for very small fields. (Actually backscatter versus field area yields a straight line when plotted on semilogarithmic graph paper.) With tiny fields, such as 1 \times 1 cm, percent backscatter is extremely small.

It was noted several years ago that, for a given field size, maximum backscatter occurs at qualities of about 0.5 to 0.8 mm copper HVL. The exact quality at which this occurs varies somewhat with field size. It is interesting and conceptually profitable to consider why such a maximum should exist.

TABLE 3. PERCENT BACKSCATTER VS. BEAM QUALITY AND FIELD SIZE⁵

Beam Quality (HVL)	Percent Backscatter for Field Area of:				
	0 cm ²	50 cm ²	100 cm ²	200 cm ²	400 cm ²
1 mm Al	0	18	20	23	25
2 mm Al	0	21	25	28	30
4 mm Al	0	26	31	35	38
1.0 mm Cu*	0	27	34	42	47
2.0 mm Cu	0	23	29	37	41.5
3.0 mm Cu	0	19.5	25	31.5	36
Cesium-137	0	3.5	4.5	5.7	6.5
Cobalt-60	0	2.0	2.6	3.7	4.6

* Orthovoltage data for closed applicators.

In very low energy photon beams there is relatively little scatter of *any* type because attenuation is primarily photoelectric. Although there is still some coherent scatter, it is readily removed before it can return to the skin because photons are effectively attenuated by tissue at low energies. In high energy photon beams there is relatively little photoelectric effect. Compton scatter is the dominant mechanism, so there are many scattered photons of substantial energy. In such beams, however, scatter tends to be primarily forward since higher energy photons are deflected on the average through smaller angles. Thus, scattered cobalt-60 rays tend to emerge traveling in a forward direction. This explains why the percent backscatter for cobalt-60 is of the order of 3 percent, which is less than one-tenth as great as that for 200 kV radiation. In the orthovoltage region the dominant attenuation mechanism is *also* Compton scatter, but the photons involved are of such low energy that many are scattered both backward and laterally. There is substantial backscatter as a result, and the observed maximum occurs in this photon energy region.

Percent Depth Dose

Depth dose tables indicate the relative central ray dose versus depth, with various size fields, for a given quality of x-rays. In addition to the depth dose, the backscatter factor or percent backscatter is also given.

DEPTH DOSE TABLES. Table 4 shows an abbreviated version of such a depth dose table to indicate the format employed. Note that each

TABLE 4. SAMPLE PERCENT DEPTH DOSE TABLE*—
3.0 MM Cu HVL, 50 CM TSD, DIAPHRAGM LIMITED

Depth in cm	Field Size—cm					
	0	5 × 5	7 × 7	10 × 10	15 × 15	20 × 20
	Percent Backscatter					
	0	14.2	18.5	23.7	29.6	33.7
0	100.0	100.0	100.0	100.0	100.0	100.0
2	68.0	86.7	89.7	92.6	95.6	96.8
4	46.4	66.2	71.4	76.7	82.1	85.7
6	32.0	49.3	54.5	60.6	67.0	71.3
8	22.0	36.2	41.1	47.2	53.7	57.9
10	15.4	26.7	30.8	36.3	42.6	46.7
12	10.7	19.4	23.0	27.7	33.3	37.2
14	7.5	14.3	17.2	21.1	25.9	29.4
16	5.3	10.5	12.8	16.0	20.2	23.2
18	3.7	7.7	9.6	12.2	15.7	18.5
20	2.6	5.6	7.0	9.4	12.3	14.5

* Fuller data given in the British Journal of Radiology, Supplement 10, from which these figures are obtained.⁵

table corresponds to a given beam quality and source-skin distance. In addition, for orthovoltage beams the type of collimator is also specified (p. 225). The table is arranged to give the percent depth dose in columns corresponding to the field size employed. The left hand column indicates the depth in centimeters below the surface. In some older depth dose data, the columns correspond to field areas without any distinction as to their shape. In the Supplement 10 data, fields are described by their specific dimensions, as either rectangles or squares. For field areas of shapes different from those specified in the headings, simple interpolations can be made; they are described in an appendix to Supplement 10. In addition to percent depth dose, percent backscatter is also given for each field size shown.

An example will serve to illustrate the use of a depth dose table. In Table 4, what are the percent backscatter for a 10 × 10 cm field and the depth dose at 6, 10, and 14 cm? The figures are seen to be, respectively, 23.7 R per 100 air-R and 60.6, 36.3, and 21.1 R per 100 skin-R. For other field areas, or intermediate depths, interpolation procedures are performed. For example, the percent depth dose for a 10 × 10 cm field at 11 cm depth is about 32.0 R per 100 skin-R.

TABLE 5. PERCENT DEPTH DOSE AT 10 CM DEPTH VS. BEAM QUALITY AND FIELD AREA—50 CM SSD⁵

Beam Quality HVL	Percent Depth Dose at 10 cm Depth for Field Area of:				
	0	50 cm ²	100 cm ²	200 cm ²	400 cm ²
1 mm Cu*	12	24	29	34	38
2 mm Cu	14	27	32.5	38	43
3 mm Cu	16	28.5	34	40	44
Cesium-137	29.5	41.5	44.5	47	49.5
Cobalt-60	38	47	50	52	55

*Orthovoltage data for *closed* applicators.

(Normally, interpolation is unnecessary in the original tables, which give fuller data.)

DEPTH DOSE VERSUS TREATMENT FACTORS. The percent depth dose of an x-ray beam is essentially its effective penetration in the body. It therefore depends on broad beam attenuation of the incident beam and divergence. The former involves beam quality (implied by the kilovoltage, wave form, and total filtration of the generator and tube) as well as field size. Divergence is determined by the treatment distance.

Contributory scatter is quite substantial in orthovoltage beams. For example, as previously indicated, approximately two-thirds of the intensity at 10 cm depth in a 10 × 10 cm field 200 kV beam results from contributory scatter.

Some examples serve to illustrate how these various factors operate. Table 5 illustrates the relationship between percent depth dose, field size, and beam quality (SSD fixed at 50 cm). Note that with very small field sizes the cobalt-60 depth dose figures are much greater than those of the 2 mm copper HVL beam. At practical field sizes (100 to 200 cm²), however, the greater contributory scatter from orthovoltage radiation tends to make the depth dose values at 10 cm depth more nearly equal.

Figure 4 compares percent depth dose curves of several beam qualities for 10 × 10 cm fields. The orthovoltage beam delivers far less dosage at 10 cm depth than either the supervoltage x-ray or cobalt-60 beams. The 4 MV linear accelerator beam gives the highest depth dose, with cobalt-60 in between. The dotted curve indicates the orthovoltage beam depth dose curve for 80 SSD; this is a decided improvement, but the percent depth dose at 10 cm depth remains only about two-thirds that of the harder beams. It is seen that beams of higher energy photons have greatest penetration, other factors being

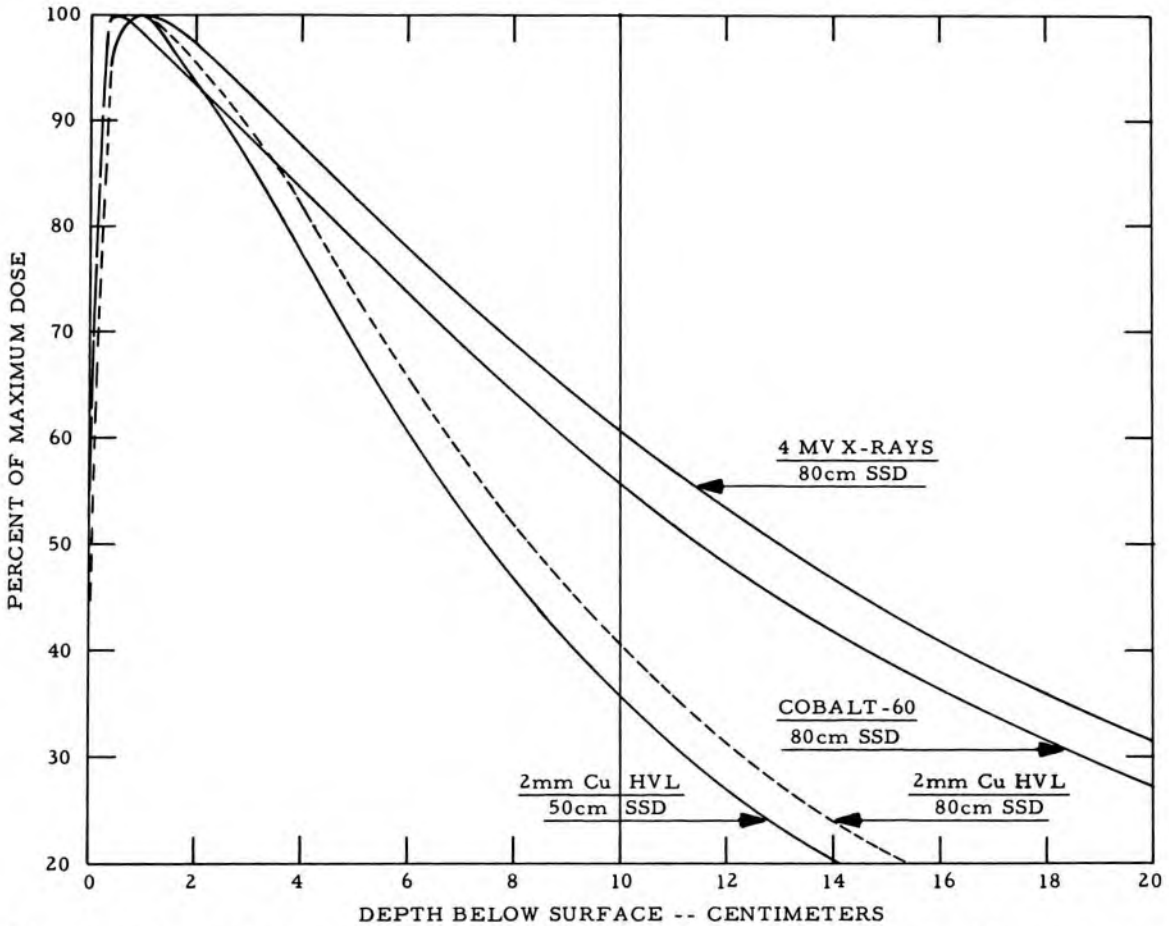


Fig. 4. Comparison of percent depth dose vs. depth curves, 10 x 10 cm field. Solid curves: lower, 2 mm Cu HVL x-rays, 50 cm SSD; middle, cobalt-60 teletherapy, 80 cm SSD; upper, 4 MV x-rays, 80 cm SSD. Dotted curve: 2 mm Cu HVL corrected to 80 cm SSD.

the same. We shall see below that supervoltage radiation has other advantages as well.

It must be stressed that SSD is a very important factor also. Note that an increase of 50 to 80 cm significantly increases the orthovoltage depth dose (36 to 40.5—over 12 percent increase). Unfortunately, increasing treatment distance greatly reduces the beam intensity, so treatments are $(80/50)^2 = 2.56$ longer at the 80 cm distance. Well-filtered orthovoltage beams have relatively low intensities, so treatment times become excessive at SSD values much greater than 50 cm. One of the practical advantages of supervoltage machines is that output is far greater at a given SSD than is attainable with orthovoltage units, making treatment at long distances feasible. As usual with x-ray equipment, this benefit is achieved at considerable financial cost.

EXAMPLES. It is interesting and useful that a hard and a soft beam of x-rays can be used in such a manner as to have exactly the

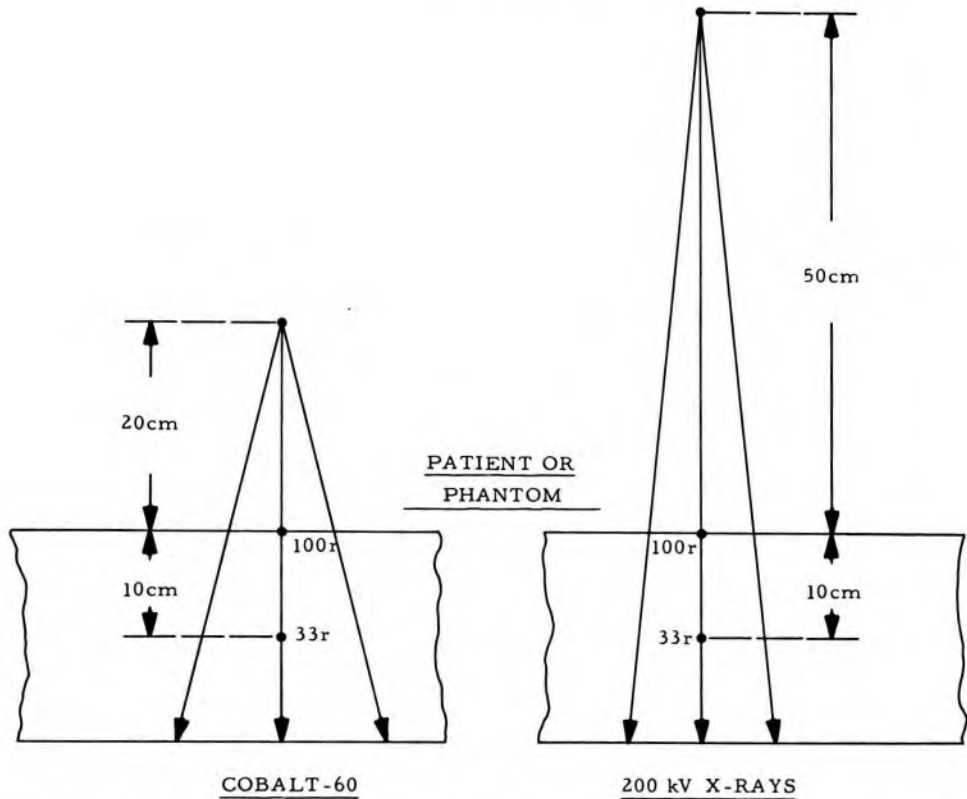


Fig. 5. Short SSD cobalt-60 teletherapy and 50 cm orthovoltage beams yield similar percent depth dose values. Cobalt-60 rays are attenuated less. This tends to produce greater depth dose; but a shorter SSD is used (20 vs. 50 cm). This tends to reduce depth dose, by greater divergence. The net effect is similar effective penetration with the two beams.

same percent depth dose curves. For example, a 200 kV x-ray beam employed at 50 cm SSD matches the penetration of a cobalt-60 beam used at 20 cm (Fig. 5). The harder beam of the cobalt machine is attenuated less than the x-ray beam and would produce a higher percent depth dose at the same SSD. Since it is employed at a shorter treatment distance, however, it diverges more than the x-ray beam. Consequently, the cobalt-60 beam loses as much effective penetration by its greater *divergence* as it gains by its greater *hardness*. Cobalt-60 radiation delivered at short SSD values is quite useful in treating relatively superficial structures around bone and cartilage, which must be irradiated with a hard beam to keep the rad dose down to reasonable levels. This is particularly important in treatment around the head and neck. In such applications one deliberately reduces beam penetration to spare deeper structures.

Another interesting application uses a hard source of radiation at short SSD for treating superficial lesions. An example is the application of radium or similar radioactive materials in plaques to

treat superficial malignant disease. This situation is the exact opposite of treatment of deep lesions, in which one desires as *high* a percent depth dose as possible. In treating superficial lesions one wishes to deliver adequate dosage to the tumor bed (e.g., at 0.5 or 1 cm depth) while at the same time delivering as *low* a dose beyond this depth as possible. Although this can also be accomplished by the use of soft x-rays at usual SSD values (about 20 cm), one can actually obtain a more rapid dosage falloff with depth using a radium plaque. In addition, the use of radium protects any underlying bone and cartilage necessarily irradiated along with the superficial lesion. As pointed out above, the absorbed dose from any soft x-rays in such materials is disproportionately high, and that from much harder radium rays practically the same as in adjacent soft tissues.

PRACTICAL ASPECTS. A few comments about the application of depth dose data are in order at this point. The appropriate depth dose tables of Supplement 10 must be employed in orthovoltage radiotherapy, or significant errors result. In general, "closed applicator" data may be used with covered cones and "diaphragm limited" data with adjustable diaphragms or uncovered cones. The rule, however, often involves some approximation.

The field shape must also be considered in using depth dose data. One should employ tables for the particular rectangular shape fields employed, or as close to these as possible. Generally speaking, a rectangular field yields a smaller percent depth dose than a circular or square field of the same area because contributory scatter is attenuated more on the average in its journey to the central ray. Such effects are much greater in orthovoltage than supervoltage beams.

In some cases, particularly in older orthovoltage machines, the actual treatment distance differs somewhat from those given in the isodose and depth dose data which are available. This should be carefully checked before initiating therapeutic procedures. Generally, if the distances are not the 50 cm given in the literature, it is most simple to obtain new cones or make other arrangements to assure that the distance is 50 cm. This at least will assure that the SSD is appropriate to the tables and isodose charts employed in computation.

If it should be desirable or necessary to treat at a somewhat different distance from that for which percent depth dose data are available, new tables can be computed, using formulae derived by Johns and others. Somewhat different formulas are employed in the region from superficial therapy up through orthovoltage than for supervoltage. These are all described in Supplement 10.

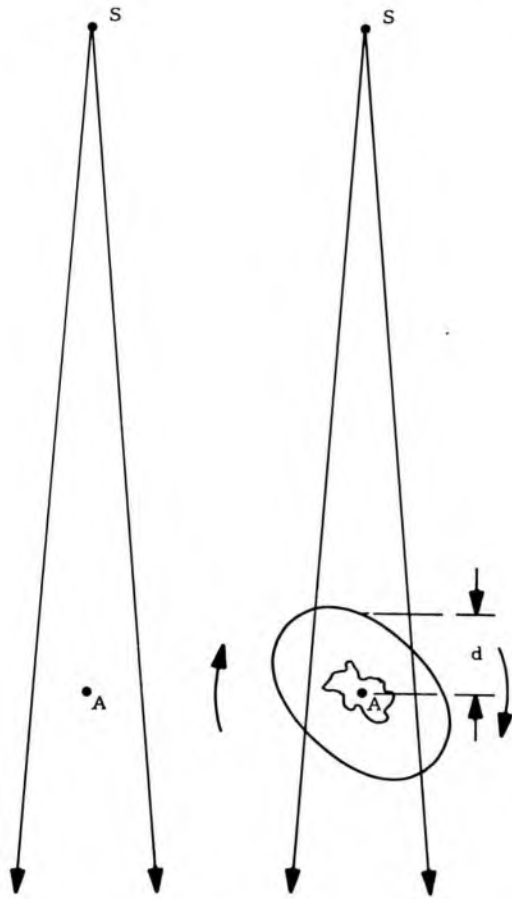


Fig. 6. Tissue-air ratio (TAR) defined. The beam output and collimation is kept constant. Left. First, the dose rate is measured at A, without patient in place. This is the *air-dose* (D_A). Right. Then, with patient in place, the new dose at A is D, the *tissue dose*, and: $TAR = D/D_A$. The D value, and hence the TAR depends on the patient's thickness d, which varies both attenuation and the SSD value.

Tissue-Air Ratio

The percent depth dose data described above involve both divergence and attenuation processes. Since divergence is involved, SSD must be specified in such data. The tissue-air ratio has been developed to permit compiling depth dose tables *independent of SSD*. Although this was originally done to simplify calculations in rotational therapy, such tables are also used for other work.

Figure 6 indicates the meaning of the term tissue-air ratio (TAR). The beam *without* the patient in place is shown on the left, *with* on the right. The tumor-air ratio is now defined as the dose in roentgens to the patient at the point in question, divided by air-R at the same location in the beam. Since the distance has not been changed, only beam attenuation quantities are involved. Thus, the TAR depends on the tissue depth and the half-value layer, which affect *narrow beam absorption*, and the field size at the location in question, which affects *contributory scatter*.

One might expect that the TAR should be substantially independent of source-tumor distance since the attenuation and scatter for a given HVL depend primarily on "d" and field size. Extensive

TABLE 6. SAMPLE TISSUE-AIR RATIOS FOR SQUARE FIELDS*—
3.0 MM Cu HVL, DIAPHRAGM LIMITED

Depth in cm	Field Size in cm					
	0	5 × 5	7 × 7	10 × 10	15 × 15	20 × 20
0	1.00	1.14	1.19	1.24	1.30	1.34
2	0.736	1.06	1.14	1.23	1.34	1.39
4	0.541	0.850	0.947	1.07	1.22	1.31
6	0.401	0.676	0.774	0.902	1.06	1.15
8	0.296	0.521	0.611	0.736	0.903	0.991
10	0.222	0.400	0.476	0.587	0.746	0.834
12	0.164	0.303	0.367	0.461	0.605	0.689
14	0.123	0.232	0.287	0.360	0.483	0.564
16	0.092	0.117	0.216	0.281	0.389	0.459
18	0.068	0.135	0.166	0.220	0.310	0.373
20	0.051	0.103	0.129	0.172	0.247	0.303

* Fuller data given in the British Journal of Radiology, Supplement 10, from which these figures are obtained.⁵

measurements have verified that this is truly the case for a substantial range of source-tumor distances.

Table 6 is a TAR depth dose table, corresponding to the same HVL and field sizes as Table 4. There is a general similarity in appearance between the tables but with three major differences. First, Table 6 makes no reference to source-tissue distance (STD) because this can be varied through a wide range without affecting the TAR values. Second, backscatter factors are not given explicitly; and finally, the field size is that of the beam *in air at the point in question*—not generally at the skin location.

Consider an example: Suppose we are interested in the dose at 10 cm depth with a 10 × 10 cm field. The TAR factor is 0.587. This means that 100 air-R at this location delivers $100 \times (0.587) = 58.7$ R to the patient. The loss of the other 41.3 R resulted from attenuation in traversing 10 cm of the patient.

Consider another example: A 10 × 10 cm field at 0 cm depth (the skin). From the table, 100 air-R delivers $100 \times (1.24) = 124$ R. This is of course to the skin. From Table 4 one would expect 123.7 R—very good agreement. This value is independent of the distance used and serves to illustrate the fact that percent backscatter depends on field size and beam quality only, but *not* on treatment distance.

A major application of tumor-air ratio data is rotational therapy. By dividing the patient into several sectors (usually 12) and adding the contribution for each of these 12 angles during the rotation, it is

relatively easy to compute the total dose at the axis during the completion rotation. (See page 000 below.) This can be done rather easily using TAR tables. Note the summation procedure is very difficult to carry out with ordinary tables because SSD, depth, and field size *all* vary continuously in rotational therapy since patients are not of circular cross section and lesions are not necessarily centrally located.

TAR data represent a more scientific approach to the computation of percent depth dose, and some consideration should be given to maintaining a constant source-tumor distance rather than SSD in treatment. Radiation therapy has been with us a long time, however, and habits are difficult to change, so fixed SSD techniques will likely be in use for some time.

ISODOSE CHARTS

We have discussed to now only central ray depth dose data. In addition one requires dosage information for other locations. Such data are presented in sets of "isodose curves," called "isodose charts."^{6,7} These charts show the dosage distribution in a plane passing through the central ray; in rectangular fields two charts are normally used, each for a plane parallel to one of the sides. For example, a 10×12 cm field has two charts: one for a beam 10 cm wide at the surface, the other 12 cm wide. Figure 7 shows an isodose chart for a 10×10 cm field orthovoltage beam (2 mm Cu HVL, 50 cm SSD, closed end applicator).⁶ Each curved line represents locations in the patient receiving equal dosage (90, 80, 70, etc. percent of the central ray skin dose). The central ray is also shown, and diagonal lines at the sides indicate the approximate geometric margin of the beam.

Isodose charts provide depth dose information for single treatment fields. They are also indispensable for figuring out the total dose distribution when several fields are combined in deep therapy. (Recall that multiple fields are always necessary to safely deliver definitive dosage to deep tumors.)

Description

In general, isodose curves are *not* flat. Thus, in Figure 7 the 50 percent isodose line crosses the 6 cm depth location at points P and Q, about 4 cm off-center. Point R on the central ray, at the same depth, receives 57 percent of maximum. Similarly, at 10 cm depth the dose falls from 32 percent at the center to 27 percent 1 cm from the margin of the beam. The decrease is even more rapid beyond this location. This effect is undesirable because radiotherapy is most

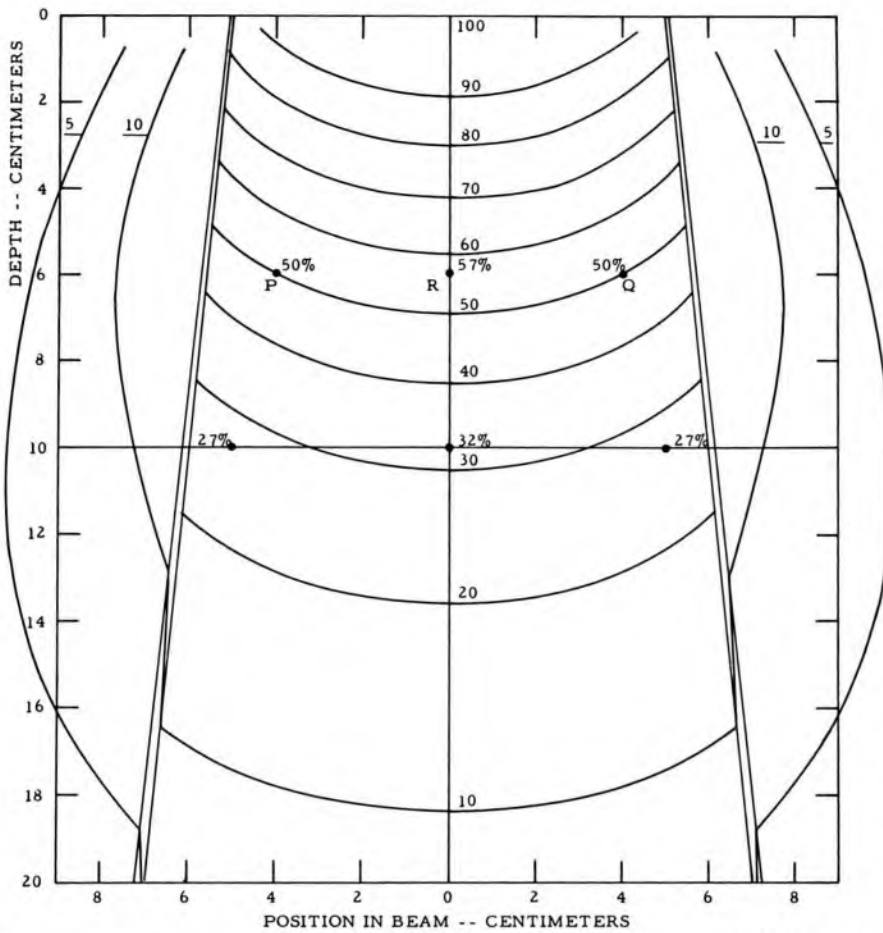


Fig. 7. Typical orthovoltage isodose chart, 2 mm Cu HVL, 50 cm SSD, 10×10 cm (closed end applicator) (see text). (Redrawn from Tsien and Cohen.⁶)

effective when uniform dosage is delivered to the tumor bed (Chap. 5).

WHY ARE ISODOSE LINES CURVED? Three processes contribute to reduced dosage near the margin of an x-ray beam. First, a ray traveling obliquely from the source travels further in both air and tissue than the central ray. Its longer total path increases *divergence*; also, attenuation is greater for the same reason. These effects, however, are relatively small compared with the second process, *contributory scatter*, which is maximum in the center and minimum at the sides. This difference arises because the central ray receives scatter from both sides through relatively short distances. Near the edge, scatter comes primarily from the *beam* side because minimum primary radiation reaches beyond the edge. Orthovoltage beams have great contributory scatter, so there is correspondingly great nonuniformity of tumor dose at depth. Supervoltage beams are scattered primarily in a *forward* direction, so there is less contributory scatter to the central ray. Consequently, isodose lines in such beams are less curved than those of orthovoltage beams.

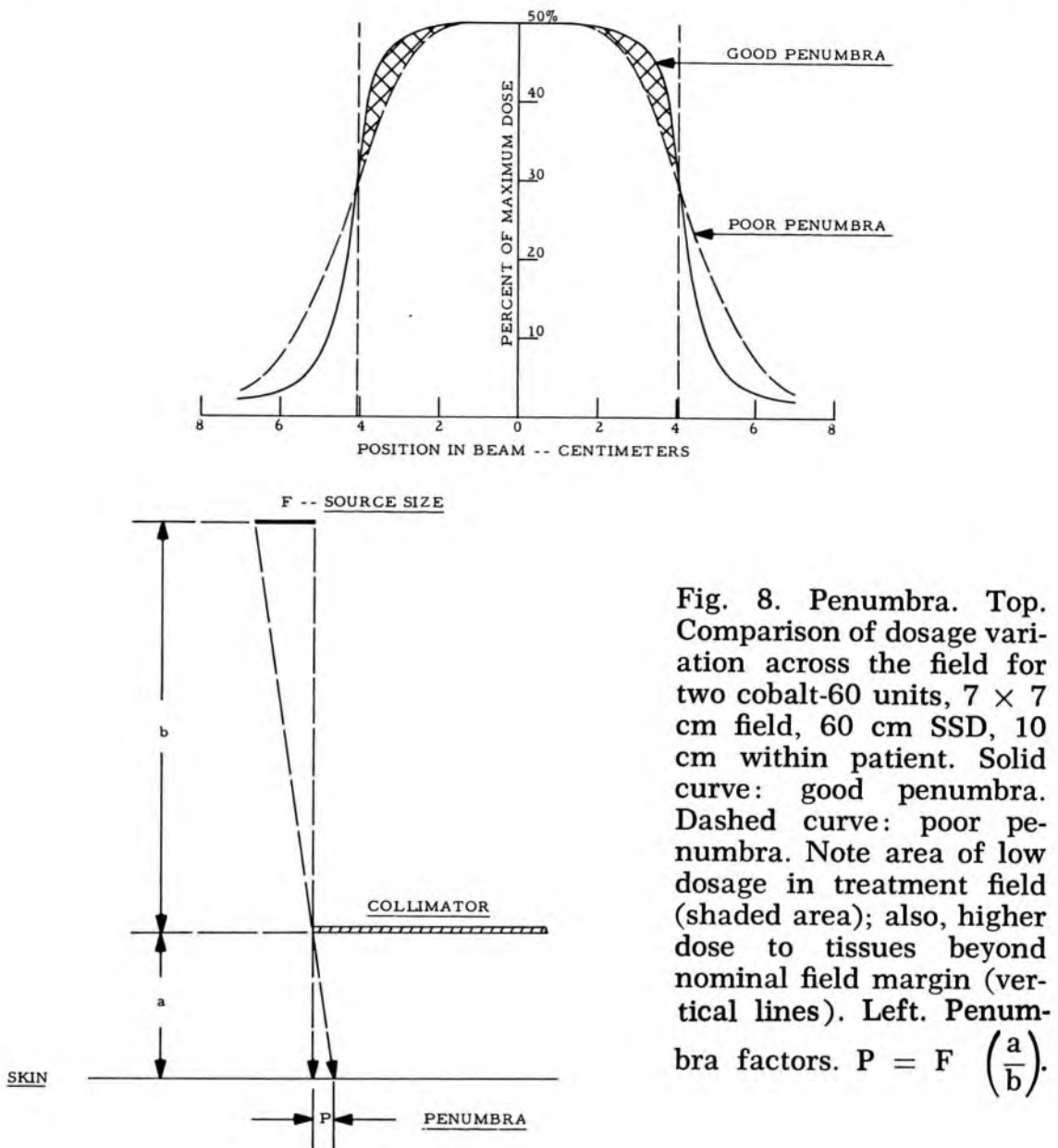


Fig. 8. Penumbra. Top. Comparison of dosage variation across the field for two cobalt-60 units, 7×7 cm field, 60 cm SSD, 10 cm within patient. Solid curve: good penumbra. Dashed curve: poor penumbra. Note area of low dosage in treatment field (shaded area); also, higher dose to tissues beyond nominal field margin (vertical lines). Left. Penumbra factors. $P = F \left(\frac{a}{b} \right)$.

Some cobalt-60 and cesium-137 teletherapy beams however vary greatly in intensity across the beam *even in air*, due to the use of very large sources and improper collimation. To illustrate, Figure 8(Top) shows the intensity distribution of two teletherapy beams, one with good and the other poor "penumbra."

We shall show in the next chapter that any source of size F produces a region of gradual intensity reduction at the sides of the beam (penumbra) given by:

$$P = \frac{a}{b} F \quad (7-3)$$

where a = collimator distance to location of interest
 b = collimator distance to source
 F = source size.

[See Figure 8(Side).] For example, consider a 3 cm cobalt-60 source, used with $a = 30$ cm, $b = 30$ cm (the SSD is 50 cm, and we are interested in a 10 cm deep tumor). Then,

$$P = \left(\frac{30}{30}\right) \times (3 \text{ cm}) = 3 \text{ cm}$$

In practice the use of round sources improves the situation slightly, but an excessive penumbra still results.

The term penumbra is often defined in radiotherapy as the separation of the 80 and 20 percent of central ray intensities in air at the desired SSD. Good penumbra is generally considered to be 1 cm or less.

EXAMPLES. Figure 9 compares five radiotherapy beams as to their dosage variation across the field at 10 cm depth in the patient. This was computed using isodose curves in references (6) and (7). All have a true 10×10 cm portal at the skin. Three are low penumbra supervoltage beams used at 80 cm SSD (D, C, and A, 22 MV betatron, 4 MV linear accelerator, and cobalt-60, respectively). For comparison, corresponding curves are shown for a poorly collimated cobalt-60 beam (B, 50 cm SSD) and an orthovoltage beam of 2 mm Cu HVL (also 50 cm SSD).

Supervoltage beams D, C, and A have relatively constant intensity from the central ray to the side, where they fall off rather rapidly.

Cobalt-60 beam B has one of the worst intensity characteristics of all and delivers considerable radiation outside the field. This is primarily because of poor penumbra and is inherent in the beam emerging from the machine. As one might expect, the orthovoltage beam (heavy curve) also delivers substantial dosage outside the treatment field because of its great lateral scatter in tissue.

A 9 cm diameter tumor receives about 99, 95, 92, 86, and 88 percent of the central ray dose at the margin, for beams D, C, A, B, and orthovoltage. Six centimeters from the central ray, healthy tissue receives 29, 15, 24, 46, and 66 percent, respectively. Thus, all well-collimated supervoltage beams irradiate the tumor uniformly and adequately spare surrounding healthy tissues. The other two beams deliver relatively inhomogeneous tumor dosage and substantially irradiate adjacent healthy tissues.

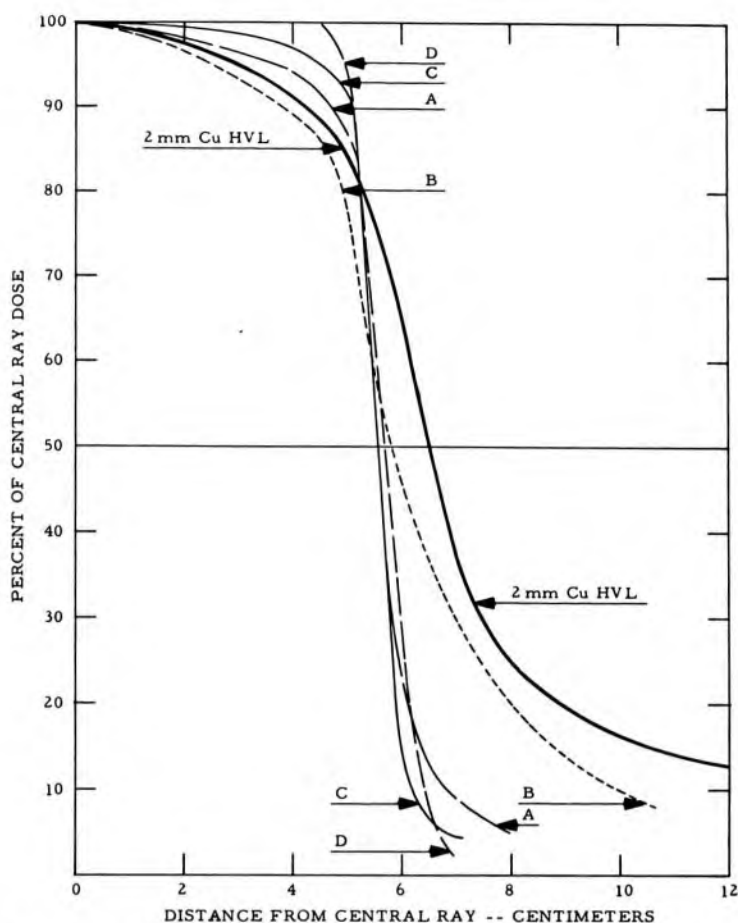


Fig. 9. How dosage varies across the beam, 10 cm deep in patient, for several treatment beams, 10 × 10 cm portal. Heavy curve, 2 mm Cu HVL, 50 cm SSD, closed cone. A—well-collimated cobalt-60 unit (80 cm SSD). B—poorly-collimated cobalt-60 unit (50 cm SSD). C—4 MV linear accelerator unit (80 cm SSD). D—22 MV betatron unit (80 cm SSD). B offers slight improvement over the orthovoltage beam, whose scatter is substantial. However, all three better supervoltage units have good penumbra, with acceptably low scatter beyond the margin of the treatment beam.

In general small cobalt-60 units tend to have excessive penumbra for deep therapy and may not be much better than orthovoltage machines in localizing treatment to the desired volume. They have been used satisfactorily, however, for deep therapy with newer small sources (1 cm) and in some less demanding treatment situations in which penumbra is not so critical.

One might then ask why large penumbra beams are used nevertheless for definitive therapy. The reason is primarily economic. A much lower activity and hence cheaper cobalt-60 source can be used with the same beam intensity at shorter treatment distances (i.e., 1,000 versus 2,560 curies for 50 versus 80 cm SSD). This would not be so bad if both "a" and "b" of (7-3) could be reduced in proportion. Unfortunately, "a" must necessarily include at least 10 cm in the

patient since the penumbra *at the tumor* is of interest. It must also include 15 to 20 cm collimator-skin separation to preserve the skin-sparing action, one of the desired benefits of supervoltage. Finally, larger diameter sources are much cheaper for the same activity because they require less nuclear reactor time. This combination makes it much more difficult to achieve good penumbra values with short SSD teletherapy machines.

It is evident that a poor penumbra cobalt-60 machine represents an initial saving in capital investment, but at the cost of less satisfactory patient treatment. As a result, high activity units are favored today over the old "hectocurie" lower activity machine.

Multiple Portal Summations

Since fields must be combined to deliver tumor-lethal dosage of radiation to deep tumors, we shall discuss how isodose charts are used to summate contributions of multiple fields.

BASIC PROCEDURE. The procedure involves the following basic steps:

1. The patient contour and anatomic cross section are determined at the level of treatment for the plane determined by the central rays of the contributory fields. This is sketched full size on translucent paper.
2. A rectangular array of points is marked on this paper, one or two cm apart. Also, relevant anatomic locations are sketched showing the tumor and vulnerable organ sites for reference purposes.
3. Isodose curves are then placed under the work sheet, one at a time, and positioned appropriately. By interpolation, the contribution from each beam to each grid point is read off and tabulated on a work sheet.
4. After interpolation from all the fields, the contributory doses are summated and noted on the grid.
5. Finally, the summated isodose curves are drawn for cardinal locations (i.e., 100, 120, 140 percent, etc. of the skin or maximum dose from any particular portal).
6. The interpolation is in general best carried out on an illuminated horizontal surface, so the anatomic cross section and isodose chart information are superimposed.

EXAMPLE. This procedure is best explained using an example. Figure 10(Top) shows two 14×10 cm cobalt-60 treatment fields aimed in opposite directions at a central pelvic lesion, with their central rays coincident (parallel-opposed portals). The pelvis is assumed 18 cm thick front-to-back, and the isodose curves are those for the

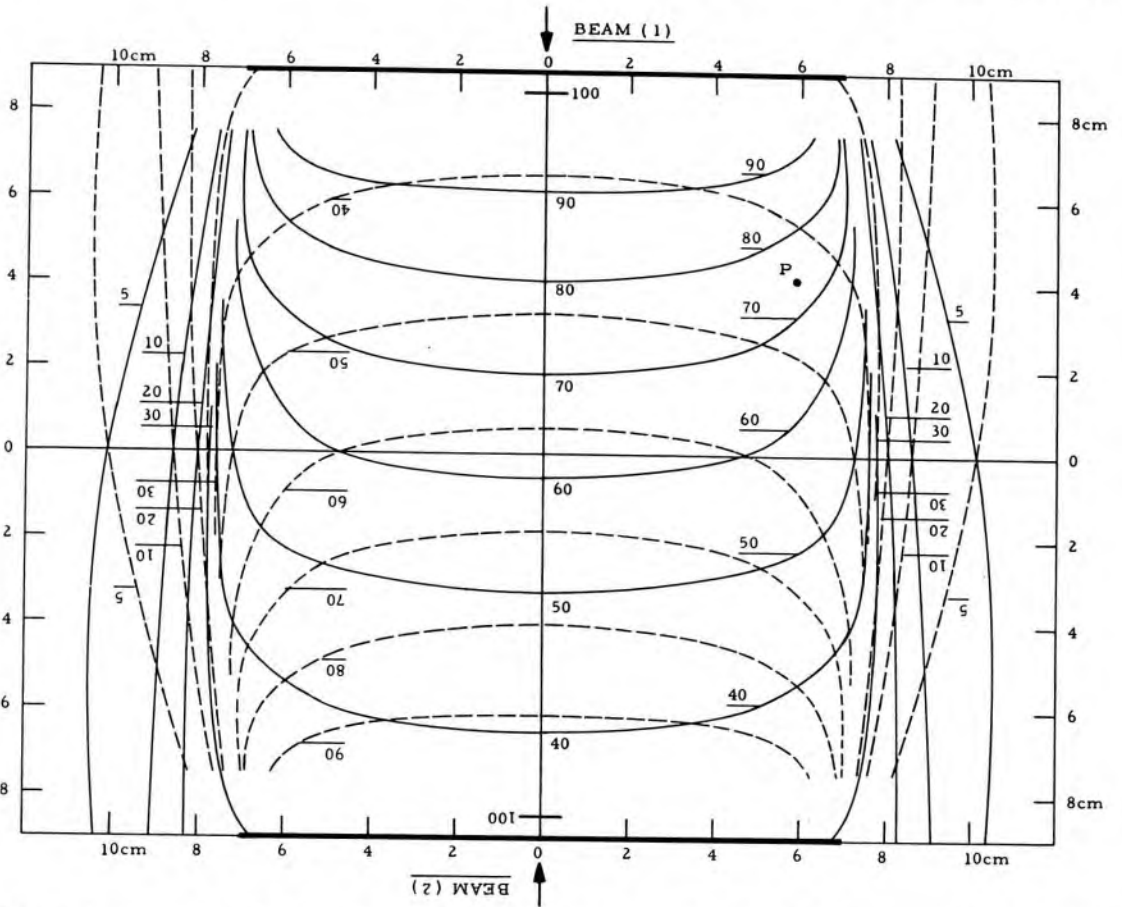
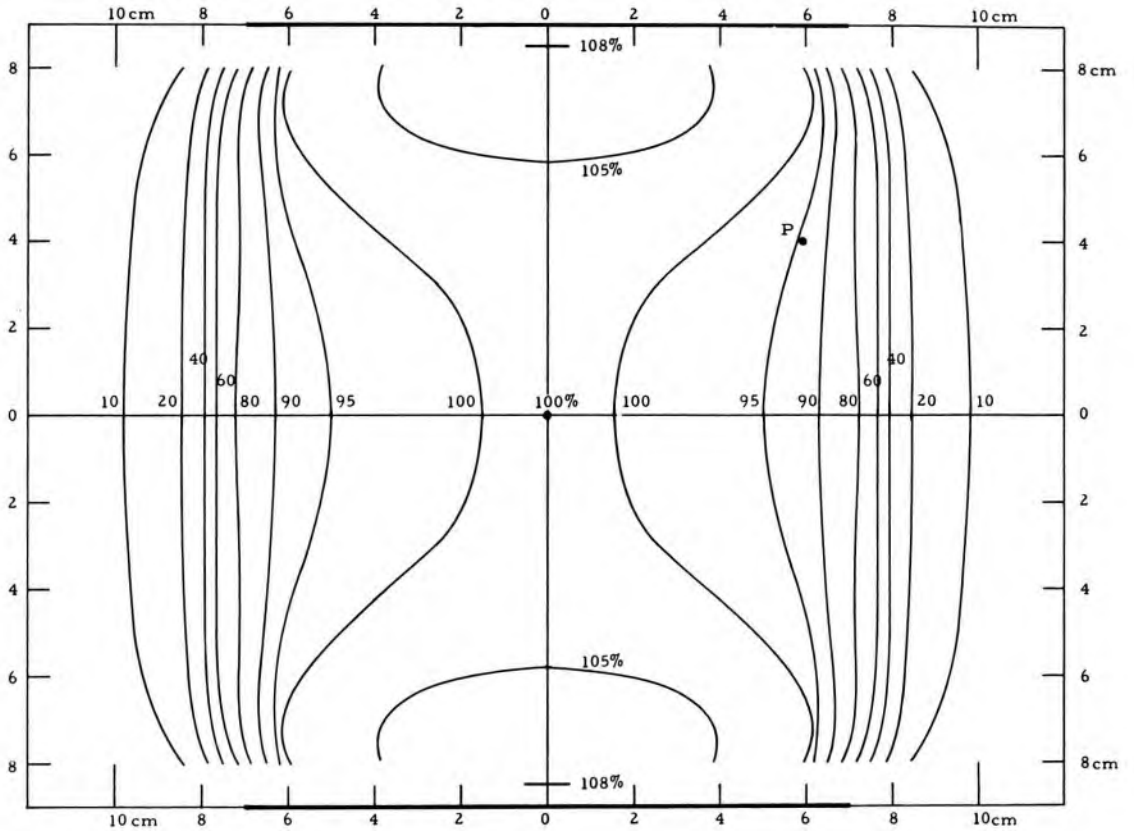


Fig. 10. (Top and Bottom). Summation of two parallel opposed cobalt-60 treatment portals (see text).



Picker C-5000 with Johns collimator. The isodose contributions of both beams are shown, those of beam (1) with solid, those of beam (2) with dashed lines. The total dose distribution is shown in Figure 10(Bottom), for 100 percent at the center of the pelvis. The direct summation yields 125 percent of the maximum of either field for this location, but the 100 percent presentation is convenient in estimating injury to areas surrounding the tumor.

The dose at point P in Figure 10(A) is 74 from beam (1), 44 from beam (2) per 100 at 0.5 cm depth from each field. The total is $74 + 44 = 118$. This corresponds to $62.5 + 62.5 = 125$ at the center. Hence the dose at P for 100 at the center is $100/125$ times this, or 94.5, the value shown in Figure 10(Bottom).

It is instructive for the reader to follow the above procedure for several points, to verify the summated curves shown in Figure 10(Bottom).

ALTERNATIVE SUMMATION METHODS.⁸ In practice, up to six fields may be combined to treat a patient. The above procedure is time-consuming and laborious, and considerable effort has been expended to reduce the burden of routine computation. Fortunately once a large number of treatment plans have been worked out one can simply modify relatively standard procedures using central ray data to establish individualized guidelines. Also, major radiotherapy training centers employ dosimetrists and involve physicist and physician trainees in this work as an important part of their training.

One of the most important reasons for expediting isodose summations is to improve treatment planning by providing rapid evaluation of alternative plans, and the best method for doing this is by computers. Isodose data are fed into the machine in appropriate form along with the tumor location, patient contour, and the location and central ray angulation of each field. The machine has previously been programmed (an elegant word for "instructed") to summate the data and in some units can print out numbers or even isodose lines. It is likely that within a few years most major radiotherapy centers will employ computer methods for routine isodose summation.

X-ray film responds to radiation, and the final density or darkness generally increases with absorbed dose. Some industrial x-ray films respond quite linearly to absorbed dose and use of film in phantoms to summate dosage would appear to be a very useful technique.⁹

All films, however, are very dependent on photon energy, as described in Chapter 6. One obvious consequence is that they cannot be used to add doses of orthovoltage and supervoltage x-rays. In addition, errors arise from this cause in any depth dose measurement, or even a single kind of beam, because scatter produces soft radiation

in the phantom to which the film responds disproportionately. The net result is generally a false *high* indicated field width in the phantom. The error involved is proportional to the beam penumbra width and can be minimal with small penumbra beams.

Supervoltage beams with small penumbra values have been usefully measured with films in Masonite phantoms simulating the patient. The results are instructive in showing general distributions and can be quite helpful as long as their basic limitation is kept in mind.

SAMPLE FIELD SUMMATIONS. A wide variety of ways exists to combine fields. In general, however, they are some form of the following:

1. Multiple fixed fields.
2. Complete (360°) or partial (less than 360°) rotation.
3. Arc rotation plus one or more fixed fields.

Special beam shaping filters, as well as special shields, may also be used. We shall now briefly describe two basic multiple field techniques.

Figure 11¹⁰ indicates a popular technique in which anterior and posterior portals are directed toward the lesion from opposite sides with the central rays overlapping ("parallel-opposed" fields). The first example shows a 3 mm Cu HVL beam at 50 cm TSD with 6×6 cm fields. The second is for a similar sized portal and 12 cm thick part, but with a cobalt-60 unit at 80 cm SSD. Note that with the ortho-

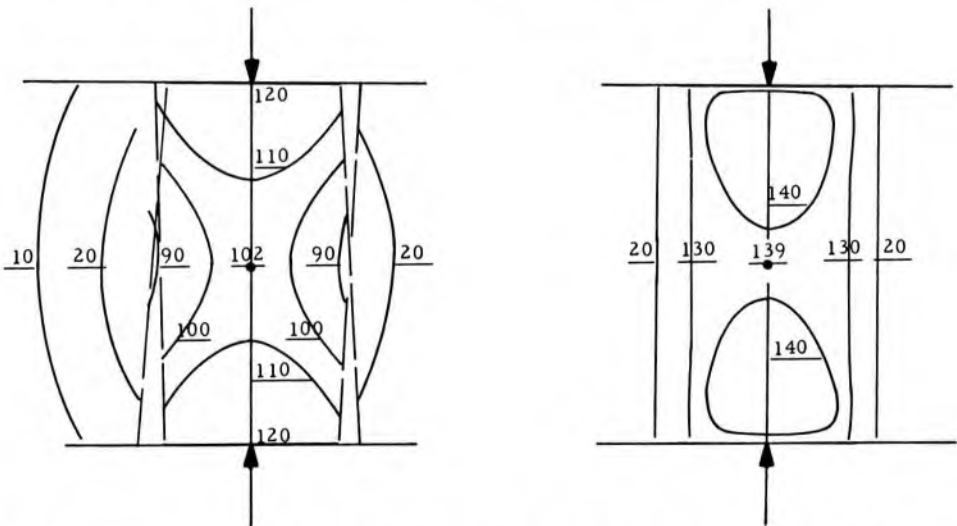


Fig. 11. Comparison of parallel-opposed portals, 6×6 cm fields, 12 cm apart. Left. 3 mm Cu HVL, 50 cm SSD. Right. Cobalt-60 teletherapy, 80 cm SSD (note this yields a much more homogeneous dose distribution in the patient). (Redrawn from Johns,¹⁰ p. 358. Courtesy of Charles C Thomas, Publisher.)

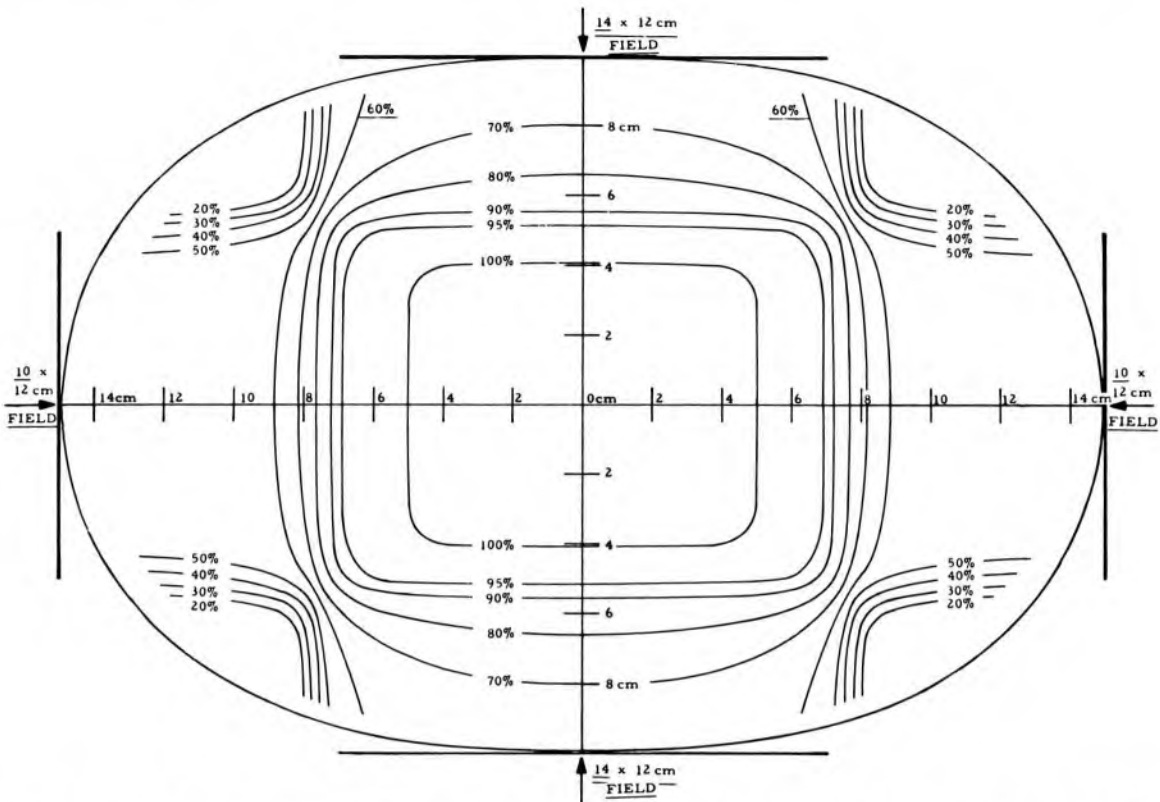


Fig. 12. Isodose summation for four-portal cobalt-60 irradiation (box technique).

voltage beam, dosage is maximum at the *surface*, falling off 18 per cent at the center; in addition, the dosage also falls off rapidly toward the side. Consequently, a deep tumor receives a relatively nonuniform dosage at 10 cm depth. The cobalt-60 dosage (Right) by contrast is relatively homogeneous both from surface to surface and side to side. There is still some narrowing of the effective treatment field at depth but far less than with orthovoltage. This illustrates the superiority of supervoltage radiation for the treatment of deep-seated lesions.

Figure 12 shows isodose curves resulting from a more elaborate procedure in which four separate fields are directed toward the tumor (the so-called "box technique") employing cobalt-60 radiation. Note that the use of the two additional lateral portals builds up the dosage received by the tumor near the lateral margins as well as increases the tumor dose significantly. With a properly planned supervoltage four portal box technique, a rectangular volume of tissue receives a substantially uniform dosage. With orthovoltage however, one normally requires at least six portals to achieve acceptable dosage homogeneity throughout a large tumor volume. Furthermore, even in this more difficult procedure, the distribution is still relatively inhomogeneous.

Rotational Therapy

Multiple portal techniques are inherently more difficult to apply than single portal techniques. Great care must be taken to accurately reproduce the individual fields in successive treatments. For many years, it has been generally agreed that it is a much simpler procedure to rotate the patient (or the machine) following an initial appropriate single setup. As the patient or beam rotates, the tumor is effectively treated from many different directions so that while the tumor receives adequate total dosage, the skin and subcutaneous tissues are spared the full dosage because of the rotation. The result is essentially the same as that in using a large number of fixed portals.

If the patient's cross section is more or less elliptical, such as in a pelvic portal, an elliptical isodose pattern is obtained for a central axis of rotation. However, the isodose curve is elliptically shaped with its major axis extending in the *anterior-posterior* direction of the patient. This results because the beam is absorbed less in the anterior and posterior portions of the treatment than during the lateral portions (Fig. 13).

Unfortunately, the advantage of convenience of rotational therapy treatment is offset by considerable complexity in dosage calculations. The central ray data may be processed quite quickly by available techniques; however, the dosage delivered at *other* locations is far more difficult to estimate. The difficulty in arriving at a convenient and satisfactory procedure for this type of estimate is indicated by the large number of alternative methods that have been proposed. Those

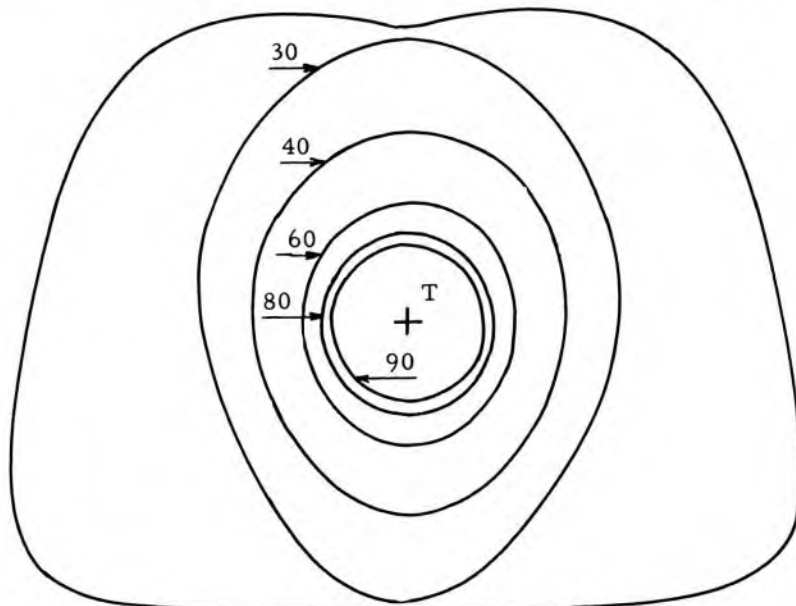


Fig. 13. Isodose summation for 360 degree rotational therapy using cobalt-60 irradiation. 15×6 cm field, 95 cm SSD. (Redrawn from Johns,¹⁰ p. 415. Courtesy of Charles C Thomas, Publishers.)

TABLE 7. SAMPLE CALCULATION OF AXIS TOTAL TAR DURING
360°-PELVIC ROTATION THERAPY 15 x 15 CM FIELD
AT AXIS (FIG. 14)

Position	d	TAR*
1	10.0	0.746
2	10.2	0.731
3	13.1	0.532
4	15.8	0.398
5	14.9	0.440
6	11.4	0.645
7	9.9	0.753
8	11.4	0.645
9	14.0	0.440
10	15.8	0.398
11	13.1	0.532
12	10.2	0.731
Total		6.991
Average TAR.		0.583

Thus, for each 100 R in air delivered at the axis rotation, the tumor receives 58.3 R at A. If 10,500 R is delivered in air during 30 days (350 R per day, 5 days per week, for 6 weeks), the tumor receives a total exposure of $(10,500)(0.583)$, or 6,120 R at the *axis of rotation*.

* For greater accuracy, original Supplement 10 data are used here, from which Table 6 was abstracted.⁵

between sample positions, the more the samples and the more accurate the result.

A sample calculation is shown in Table 7 for a pelvic rotation treatment (Fig. 14), using TAR data of Table 6.

Pitfalls in Applying Depth Dose Data

Many errors can arise in implementing even the best treatment plan. These must be minimized or taken into account to avoid jeopardizing the therapeutic result. There are three general types resulting from irradiation procedures, equipment, and limitations of depth dose data. We shall now consider each specifically.

PROCEDURAL ERRORS. These arise from uncertainties in tumor location and x-ray exposure during treatment.

The tumor site may move significantly during therapy. Visceral organs can shift substantially when the patient is moved from the

erect to supine, or even supine to prone position. "Isocentric mount" supervoltage and cobalt-60 machines are therefore very useful because they permit changing beam angulation without shifting the patient's position on the treatment couch. During a complete course of radiotherapy, radiation response can alter tumor locations as well as size, so localization roentgenograms must be repeated at appropriate intervals. Such films are especially useful when marking devices are used, such as metal clips put in place at surgery and contrast media inserted into appropriate body cavities.

The output of radiotherapy machines can vary substantially. Even cobalt-60 machine output follows the radioactive source decay of about 1.1 percent per month. X-ray generator output can fluctuate much more widely, and complete calibration is generally required at least twice a year. Checks of machine output constancy can be carried out quickly. Weekly measurements are recommended to detect serious changes promptly enough to correct the treatment plan in response to gross output changes.

Of course, there is no substitute for proper day-to-day beam direction, with correct SSD, angulation, size, and centering of all fields. Inappropriate and damaged filters, collimators, and cones can be avoided by careful procedures and conscientious and well-trained personnel.

EQUIPMENT. Even with the greatest procedural care one finds basic limitations still existing in both orthovoltage and supervoltage machines.

We have already referred to the fact that covered cones yield different percent depth dose data from uncovered cones or diaphragm-limited beams. In general, the machine employed must closely simulate the SSD, HVL, and scatter conditions under which depth dose data were measured. Published isodose curves are generally symmetric about the central ray. Actual orthovoltage beams are reasonably symmetric along a line at right angles to the tube axis but show marked asymmetry along a line *paralleling* the tube axis because tube targets are of the "reflection" type, in which x-rays emerge from the surface entered by the electron beam. These introduce a small error in isodose curve application, which increases with field size.¹¹

Supervoltage quality radiation sources also present their own problems. Cobalt-60 beams in general have larger and more variable penumbra values than x-ray beams. This can create some ambiguity as to the meaning of field size; the separation of 90, 50, and 20 percent of central ray air-R intensity have all been employed with some resulting confusion. Measurements on several commercial machines have shown "10 × 10 cm" field sizes ranged in values from 9.1 to 11.3 cm

on a side when the 50 percent intensity line separation is taken as field size.¹² (This convention is now quite general.)

It is evident that even equal field size beams with different penumbra have different isodose characteristics (Fig. 8).

Commercial 1 and 2 MV supervoltage machines are relatively stable and reliable in operation. Linear accelerators and betatrons operate from 4 MV through 35 MV; they are all more complex units with their own practical problems. One of the most serious noted in past use is asymmetry of beam intensity across the field under certain conditions. This can arise when the central ray is misaligned with the "beam-flattening filter," which must be employed at 4 MeV and higher electron beam energies (p. 251). The central ray can shift location on the target during warmup and machine orientation and when electronic components are replaced. Newer machines control this difficulty by automatically restoring alignment electronically. These higher energy machines, however, must be checked regularly for beam asymmetry to verify the proper functioning of the alignment devices.

BASIC LIMITATION OF DEPTH DOSE DATA. Published isodose data are in general applicable only to machines and factors similar to those of the original measurements. Errors in field size are particularly serious in their effect, but penumbra must also be matched in cobalt-60 units.

More fundamentally, all isodose data involve four basic assumptions. We shall now consider their nature and effect.

1. *Tissue homogeneity.* Most published data have been obtained using water phantoms. Although water simulates soft tissues reasonably well, bone and lung require appropriate corrections. Actual roentgen dose at depth is greater in lung, less in bone, than predicted from isodose data. Ignoring these corrections can result in significant errors, particularly in thoracic irradiation.

Table 8 shows air and bone corrections factors for beams of various qualities.¹³ Two illustrative examples are given in the table.

To obtain full, corrected isodose distributions requires complex procedures, and corrections are best applied by physicists working closely with the therapist, employing accurate cross-sectional anatomic drawings.

2. *Adequate scatter.* Water tank phantoms are quite large, so contributory scatter is nearly maximum. This introduces no problems in computing torso dosage, but slight discrepancies

TABLE 8. APPROXIMATE FACTORS TO CORRECT FOR
 ATYPICAL ATTENUATION OF TREATMENT BEAMS
 TRAVERSING AIR OR BONE

Beam Quality	Increased Transmission* from 5 to 8 cm of Air	Bone Shielding Reduction— Thickness of:†		
		1 cm	2 cm	3 cm
2 to 4 mm Cu HVL	1.40	0.90	0.80	0.70
Cobalt-60	1.20	0.96	0.92	0.88
4 MV	1.15	0.97	0.94	0.91
20 MV	1.10	0.97	0.94	0.91

Two Examples:

1. **Cobalt-60 treatment**—80 cm SSD, 10×10 cm. Tables give 48.1% at 12 cm depth in water. With air in lung, dose is (1.20)(48.1%) or 57.7%—much higher.
2. **1 cm thick bone in field**—6,000 r tumor dose for water alone.
 - A. 200 kV beam: actual dose = (0.9)(6,000) = 5,400 rads.
 - B. Cobalt-60 beam: actual dose = (0.96)(6,000) = 5,760 rads.

*Data from N.B.S. Handbook 87, p. 25.¹³

†Estimated from several sources.

may arise in applying such full-scatter data to irradiation of the head, neck, and extremities.

3. *Patient curvature and beam nonperpendicularity.*¹⁴ Phantoms are almost always flat rather than curved like patients. In orthovoltage work, unity density rigid materials are often placed over body surfaces to yield a flat surface perpendicular to the beam central ray. Such materials are called “bolus.” In addition to facilitating isodose summations, these devices when rigid facilitate reproducibility of field placement. In some cases departures from flatness are small, and bolus can be omitted without great error. In supervoltage work, bolus cannot be placed on the skin, and metal “wedge filters” are employed. (These are discussed below.)

In general, fields are aimed nonperpendicularly at the tumor. Bolus and wedge filter techniques are convenient. Computational techniques are also available to estimate isodose curves for the particular patient contour and angulation. These are the “effective SSD method” and the “attenuation method.” Simple and ingenious procedures are involved which are beyond the scope of this presentation, and the reader is referred to the references.

4. *Beam symmetry about the central ray.* We have already mentioned asymmetry in all reflection target (orthovoltage) beams and the possibility of asymmetry from misalignment of the electron beam with the field compensation filter (4 MV

and higher supervoltage). No such problem arises in cobalt-60 and lower voltage supervoltage machines (1 and 2 MV).

INTEGRAL OR VOLUME DOSE

The previous sections have discussed evaluation of absorbed dose (rads) in tumor and other tissue. This has essentially involved the distribution of local ion density in and around tumor. Volume dose summates all these local effects to provide a rough estimate of total trauma to the organism and becomes quite important in estimating the likely radiation sickness during radiation therapy or other situations of substantial irradiation (nuclear accident, warfare, etc.).¹⁵

Calculation

Several attempts have been made to evaluate integral dose. We shall consider Mayneord's relatively simple formula because it provides a practical indication of what is desired. The simplified formula is given below:

$$D_v = AD_o d_{1/2} \left(1 + \frac{2.88d_{1/2}}{f} \right) \quad (7-4)$$

The volume dose is given in gram rads. A gram rad is, of course, 100 ergs, from the basic definitions of the term rad.

Let us now consider the logic of this formula. "A" refers to the area of the field in cm². It is obvious that volume dose increases with the volume of tissue irradiated, so this is consistent with common sense. Also, D_o is the dose delivered to the surface (or the maximum dose, more generally) of the patient in rads, or, more correctly, the surface exposure in roentgens (including backscatter) multiplied by \bar{f} . Here too the volume dose may be expected to be proportional to this term as well. d_{1/2} is the 50 percent central ray depth location in cm beneath the skin, for the beam used. It characterizes the penetration of the x-rays; the greater d_{1/2}, the greater the dosage at all levels below the skin and consequently the greater the volume dose. For example, Figure 15 compares orthovoltage and cobalt-60 beams having the same treatment distance and superficial dosage values. Note dosage at any depth in the cobalt-60 beam is always higher than that with orthovoltage. Correspondingly the d_{1/2} is 10 cm with cobalt-60 rays and only 7.5 cm with orthovoltage. It is evident that the cobalt-60 with higher d_{1/2} has both consistently higher dosage throughout the patient and a correspondingly higher volume dose for the same skin dose, so this checks out.

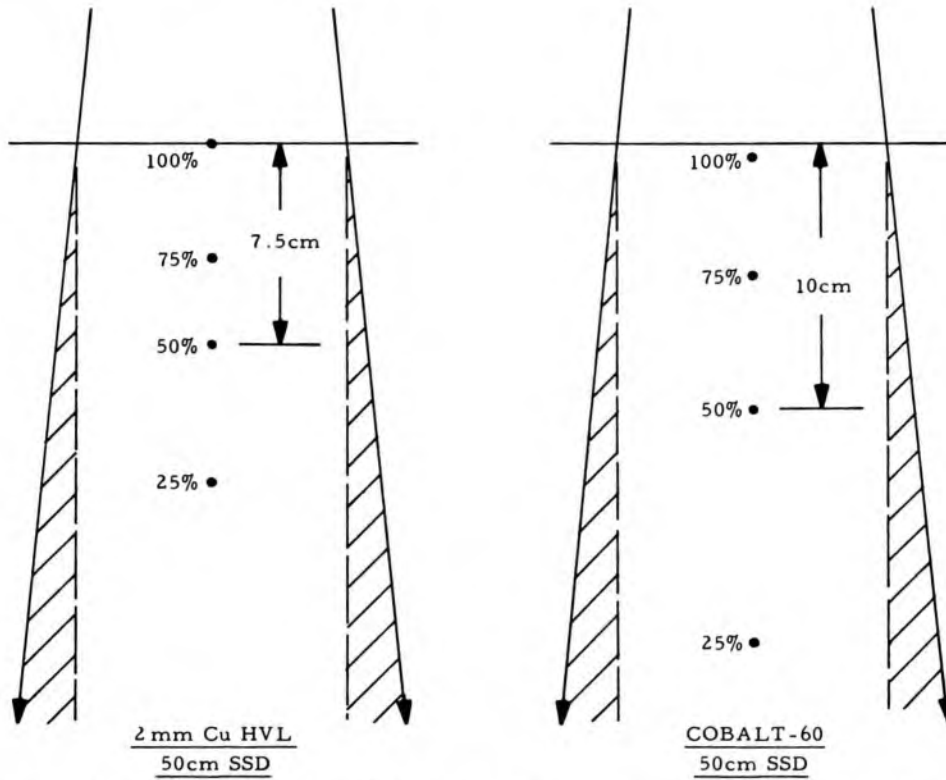


Fig. 15. Orthovoltage and cobalt-60 beams compared as to volume dose, SSD = 50 cm, 10 × 10 cm field both curves. Note that $d_{1/2}$ is 7.5 cm for the 2 mm Cu HVL vs. 10 cm for the cobalt-60 beam. Correspondingly, the dose at any given depth is higher for the cobalt-60 than for the orthovoltage beam. Shaded area represents tissue irradiated at depth beyond 10 × 10 cm field size, due to beam divergence.

The second bracket tries to accommodate the fact that if one treats at a closer distance the beam spreads out to a greater extent (f is the SSD in cm). The tissue included in the shaded areas of Figure 15 is hence also treated because of beam divergence.

Equation (7-4) gives the total volume dose for an assumed zero dosage at the exit portal; this in general is not realistic, and a simple correction is made to eliminate this limitation. The factor is $\{1 - e^{-.693d/d_{1/2}}\}$, in which d is the patient thickness.

We shall now briefly consider an example and then comment on the practical application of the formula.

Consider the volume dose from the four-portal technique such as that shown in Figure 12. Table 9 illustrates the volume dose calculation, yielding a total of 8.6 megagram rads.

Significance

What is the significance of such a number? Two limitations exist. First, the formula is only an approximation to the actual energy dissipated in the patient. It assumes flat isodose curves and zero scatter

TABLE 9. SAMPLE VOLUME DOSE CALCULATION—
FOUR-PORTAL PELVIC TREATMENT (FIG. 11, $f = 80$ CM)

Item	Field 1 14 × 12 cm	Field 2 10 × 12 cm	Field 3 14 × 12 cm	Field 4 10 × 12 cm
A	168 cm ²	120 cm ²	168 cm ²	120 cm ²
D ₀	3,000	3,000	3,000	3,000
d _{1/2}	12 cm	11.5 cm	12 cm	11.5 cm
AD ₀ d _{1/2}	6,050,000	4,140,000	6,050,000	4,140,000
f	80 cm	80 cm	80 cm	80 cm
2.88d _{1/2}	34.6	33.1	34.6	33.1
$\left(1 + 2.88 \frac{d_{1/2}}{f}\right)$	1.432	1.414	1.432	1.414
d	20 cm	30 cm	20 cm	30 cm
d/d _{1/2}	0.6	0.383	0.6	0.383
$\left(.693 \frac{d}{d_{1/2}}\right)$	0.416	0.265	0.416	0.265
$\left(e^{-.693d/d_{1/2}}\right)$	0.659	0.767	0.659	0.767
Exit Dose Factor	0.341	0.232	0.341	0.232
Volume Dose	2,950,000 gram rads	1,360,000 gram rads	2,950,000 gram rads	1,360,000 gram rads

Total volume dose is 8,620,000 gram rads, or 8.6 megagram rads.

outside the treatment field as well as uniform patient composition. Errors from the first two assumptions tend to compensate, but significant errors can result in thorax volume dose calculations. Second and more fundamentally, the biologic trauma corresponds not only to the total energy in ergs dissipated in the patient but to many subtle and complex biologic factors as well.

Consequently, the concept and calculation of volume dose should be kept in some perspective. It is literally neither necessary nor feasible to compute the volume dose to much better than 10 percent accuracy. Even with a ± 20 percent error, however, the estimate of the volume dose in megagram-rads provides a very useful indication of likely biologic injury to the organism.

SUPERVOLTAGE RADIATION

Although some early workers had unreasonable expectations for supervoltage radiation, it quickly became evident that high energy photons have no unique therapeutic properties. Their greater penetra-

tion and skin sparing have nevertheless encouraged radiotherapists to treat deep lesions more aggressively than formerly. When treatment has been carried out with good planning and management, results have been quite encouraging, and it is generally accepted that supervoltage radiation has greatly extended the scope of definitive radiotherapy.

Unjustified criticisms were raised against supervoltage therapy during the 1950's in reaction to some initial reports of radiation injury to patients from early experimental and improper usage as well as the inevitable, legitimately higher rates of complications from more definitive therapy. The dust, however, has long since settled after more than two decades of clinical trial, and supervoltage is now almost universally accepted as superior to orthovoltage x-rays for treatment of deep tumors. In addition, some higher energy machines offer the option of electron beam therapy, an exciting new tool. More remote but possibly useful modalities are neutron and proton sources for high LET radiation therapy.

In this section we summarize the major advantages of supervoltage x-ray radiation and discuss special beam-shaping filters useful in supervoltage therapy.

Comparison of Supervoltage and Orthovoltage Radiation

A summary of the advantages of supervoltage over orthovoltage for deep lesions treatment is given in Table 10.

EFFECTIVE PERCENT DEPTH DOSE. Possibly the most important advantage is greater percent depth dose, arising from two factors. First, the radiation itself is much harder and therefore more penetrating. For example, a cobalt-60 beam yields approximately 40 percent more dosage at 10 cm depth than a corresponding orthovoltage beam, for a 12×12 cm field and 50 cm SSD in both cases. In addition, supervoltage beams may generally be used with practical treatment times at distances of 80 to 100 cm rather than 50 cm, if one is willing to spend the additional sum for a stronger source and more effectively collimated beam. At 80 cm SSD this results in a 13.5 percent additional dose at 10 cm depth because of reduced beam divergence at the greater treatment distance.

Of course, a rad from supervoltage is only about 0.85 times as biologically effective as a rad from orthovoltage, so part of the advantage of higher penetration is lost. One may combine these three factors, yielding a total factor of increase in biologically effective dose at 10 cm depth, for the same superficial dose, of: $1.135 \times 1.40 \times 0.85 = 1.35$. Thus, one can in practice deliver roughly a third again as much effective dosage to a 10 cm deep tumor with supervoltage as

TABLE 10. COMPARISON OF ORTHOVOLTAGE AND
SUPERVOLTAGE*
RADIATION THERAPY MODALITIES (see text)

Item	Comment
1. Penetration—for typical portals	About 35% greater effective tumor dose at 10 cm depth with supervoltage. Factors: A. Better distance— 80 vs. 50 cm 1.135 B. Harder beam 1.40 C. Lower biologic effective- ness 0.85 <hr/> Total (product) 1.35
2. Uniformity of dosage across tumor	Better with supervoltage, especially at higher energies with beam flattening filters (Fig. 9)
3. Volume dose to patient, for given lesion size.	Lower with supervoltage because side scatter is less.
4. Bone rad dose for a given roentgen dose	Much lower with supervoltage; where orthovoltage must be used, heavy filtration desirable.
5. Ease of calculation of corrections for bone and lung absorption, oblique beam incidence, etc.	Absorption effects much smaller for supervoltage. Isodose curve corrections also relatively simple.
6. Skin sparing effect (subcutaneous vs. surface rad dosage)	Provided by supervoltage only. (However, exit and subcutaneous injury still occur.)

* We restrict this discussion to larger, well-collimated, high output machines for which the full benefits of supervoltage are obtained. Figures are for cobalt-60; situation is slightly more favorable for 4 and 6 MV linear accelerators and 22 to 35 MV betatrons.

orthovoltage! The increase is even greater with more penetrating beams like 6 MV and 22 MV x-rays. For example, consider the treatment of a moderately heavy woman for carcinoma of the cervix using external beams. With orthovoltage a maximum tumor dose of the order of 4,000 rads might be delivered without exceeding skin tolerance; with supervoltage something of the order of 5,400 rads could be delivered safely. As a result, one can adequately sterilize the pelvic lymph nodes with much less radium exposure than would be required were orthovoltage beams used, with consequently a greatly reduced probability of complications from the radium therapy.

ISODOSE CURVE FLATNESS. A second advantage of supervoltage beams is their flatter isodose lines (Fig. 9), an advantage most pro-

nounced in betatron beams but still substantial in 4 and 6 MV linear accelerators. Even larger cobalt-60 teletherapy and 1 and 2 MV x-ray machine isodose lines are much flatter than those of orthovoltage beams. This characteristic is very valuable in assuring uniformity of tumor irradiation.

VOLUME DOSE. Volume dose for a given delivered tumor dose is lower in supervoltage than orthovoltage therapy because lateral and backscatter in the patient is much less (Fig. 9). Of course, there is still much scatter in supervoltage beams but mostly in the *forward* direction.

BONE SPARING. Supervoltage radiation is less injurious to bone than orthovoltage, for the same roentgen dose delivered (p. 217). Even a well-filtered 250 kV x-ray beam contains considerable amounts of low energy photons (50 to 100 keV). These may result in at least 50 percent higher rad dosage to bone cells for the same roentgen dose than supervoltage. Use of a poorly filtered 200 kV x-ray beam, such as is sometimes employed to increase beam intensity, could actually double the rad dose delivered to bone cells. The production of osteonecrosis undoubtedly involves many other factors in addition to the rad dosage to the bone involved. The consequences of this complication, however, are sufficiently unpleasant that it seems prudent to preferably employ supervoltage in treatment to high dosage levels near bone, and to at least use very heavily filtered orthovoltage x-ray beams when supervoltage is not available.

CORRECTIONS. A fifth advantage of supervoltage, ease of corrections for patient inhomogeneity and curvature, arises incidentally from its greater penetration and flatter isodose lines. With supervoltage, corrections for bone and air absorption are rather easily made (Table 8) since tissue absorption is relatively small. Corrections for oblique beam incidence and body curvature are also rather simply made. This is of great importance in permitting simpler correction procedures and encouraging their more universal application and incorporation in practical standardized computer programs.

SKIN SPARING. The skin sparing effect has received much coverage in the literature. It arises primarily because electron equilibrium is not normally achieved at the skin surface with high energy x-ray beams. (See Fig. 6, Chap. 6.) The maximum usually occurs at 3.5, 5, and 15 mm below the skin surface for 2 MV supervoltage, cobalt-60, and 6 MV supervoltage rays, respectively. As a result, rad dosage to skin capillaries ranges from about 20 to 60 percent of that delivered to the maximum dose location, for properly administered supervoltage treatment.

Skin sparing can be jeopardized by the presence of any material placed on or near the skin (dressings, collimator block supports, etc.). In general, if material is placed nearer than 15 or 20 cm to the skin, some of its electrons are driven by Compton interaction into the skin; under unfavorable circumstances they can increase the skin rad dose considerably. When objects must be placed in the beam, special metal filters may be inserted to absorb electrons. Such "electron contamination filters" are usually made of intermediate atomic number materials like brass or steel, which themselves produce less electron contamination than either higher or lower Z materials. If at all possible, however, it is best to avoid any material in the beam altogether.

The practical effect of the skin sparing action is to shift injury from the skin to subcutaneous tissues. Although subcutaneous injury can be quite serious to the patient, it is in many cases preferable because intact skin protects the underlying tissues from infection. It must be stressed, however, that skin sparing results in practically no warning to the therapist that he is delivering an excessive dosage to deeper structures. This must be borne in mind by the therapist accustomed to orthovoltage when he first starts employing supervoltage, to avoid being trapped by habit into administering a harmful overdosage.

High Energy X-ray Machines

We have mentioned that high energy machines operating from 4 through 35 MV have distinct advantages, and, shall now comment briefly on linear accelerator and betatron units. (See Chapter 16.)

These machines all provide relatively flat isodose curves, a result achieved by the use of special beam compensation filters (see below). Such filters can be employed on these machines because considerable beam intensity is available, so use of the filters does not result in unduly prolonged treatment times. (Such filters selectively absorb central ray radiation, thereby flattening isodose lines.)

Betatron and linear accelerator machines have both been available for only a few years, and final comments cannot yet be made regarding their relatively clinical value. It may be of some interest, nevertheless, to mention some of the practical advantages and disadvantages of currently available units.

BETATRON. The betatron as most generally clinically employed operates in the range of 22 to 35 MV. Isodose curves are remarkably flat, with negligible penumbra. The x-rays, however, are extremely penetrating and therefore it is more difficult to restrict them to the desired treatment portal size. Extremely great thicknesses of lead are required in collimators to attenuate such very high energy beams (of more than 6 inches of lead). This poses manipulation problems when

large cones must be removed and new cones inserted for different treatment portals. This type of mechanical problem can of course be solved by appropriate engineering designs. Perhaps more serious is the relatively low output at larger fields of the more common betatron units, which operate at energies of about 24 MeV. This limits the number of patients who can be treated in a working day.

The betatron, however, does have the option of an electron beam of very high energy. Although the clinical benefits of such a beam have not yet been completely evaluated, they do have potential benefits of delivering their dose without significant irradiation beyond an adjustable depth within a patient. Thus, it is possible to irradiate deep lesions while sparing the tissue beyond them with betatron electron beams.¹⁶

It must be stressed that very interesting work is in progress on electron beam therapy and the physics is reasonably well worked out, but many practical problems still exist. Treatment planning and dosimetry is currently much more complex than that of photon beam therapy. The problems become especially complex in the thorax and head, where bone and air absorption discontinuities produce much greater dosage variations than those in x-ray therapy.

LINEAR ACCELERATOR. Linear accelerators commercially available so far for radiation therapy have been limited to the region of 6 MV x-rays and 8 MeV electrons. The clinical application of 8 MeV electrons is somewhat limited by their low penetration in tissue. However, 6 MV or even 4 MV x-rays with appropriate compensating filters yield quite useful, relatively flat isodose curves. Furthermore, the x-ray output of such units is very high, of the order of 200 to 400 rads per minute at a distance of 100 cm at 6 MV. As a result of these more practical considerations, there has thus far been a greater acceptance of linear accelerators than betatrons for radiation therapy.

It must be stressed that both types of machines have been in operation an insufficient period of time in major centers for evaluations to be conclusive. Moreover, design developments are continually being made, so final judgment must be reserved.

Beam Shaping Filters

Three different types of special filters are employed with supervoltage machines: beam flattening, wedge, and field compensation filters. The first is used in the machine head to obtain flat isodose curves. The second two are inserted in the beam after it leaves the machine. They are used to modify the uniform beam intensity distribution to achieve desired dose distribution within the patient for particular treatment problems.

BEAM FLATTENING FILTERS. When electrons collide with a target to produce bremsstrahlung x-rays, the intensity and direction is complicated by target x-ray attenuation and inclination to the electron beam. A detailed discussion will not be presented here; however, high energy electron beams generally produce x-rays with maximum intensity in a *forward* direction. At energies of 1 MeV and greater this effect is so pronounced that tubes employ "transmission" targets to increase x-ray output. These are usually thin tungsten or gold discs struck by electrons on the tube side with x-rays emerging in the same direction on the other side (Chap. 16).

At higher energies this directionality becomes very pronounced, and the beam intensity falls off rapidly to the side of the central ray. For example, 10° either side of the central ray the beam intensity is down about 50 percent for x-rays produced by a 10 MeV electron beam! As a result, there is a marked curvature of isodose lines with uncorrected x-ray beams produced at 4 to 25 MeV (dotted curves of Fig. 16). All linear accelerator and betatron machines for this reason employ roughly conically shaped filters near the x-ray tube to flatten the beam. They do this by selectively absorbing radiation from the central ray. The result is shown in the solid curves of Figure 16. The final isodose lines are easily made very flat in high energy betatrons and relatively flat in lower energy machines.¹⁷

One might ask why these filters are not used to flatten cobalt-60 and 1 and 2 MV supervoltage machine isodose curves. The reason lies in the difficulty of obtaining high roentgen output at these lower operating photon energies without excessive cost. Any filter functions by removal of photons, so it is impractical to use flattening filters on these machines when output is limited. Even 4 and 6 MV machines' beams are not in practice flattened to the extent theoretically possible, to assure a high output.

One further point: the larger the beam angle θ , the greater the attenuation of the central ray required to equalize the beam intensity. For this reason, betatron output can be quite high for small fields but falls off rapidly with larger fields.

We have already mentioned that beam flattening filters must be precisely aligned with the central ray, or oblique isodose lines result in both air and tissue. Mechanical stress as well as machine electrical fluctuations may theoretically affect this alignment, and linear accelerators and betatrons must be checked regularly to assure proper alignment.

WEDGE FILTERS.^{18, 19} These units are made of lead, brass, and similar materials. They are inserted in the x-ray beam to deliberately distort the isodose curve shapes, rendering them oblique. Such filters

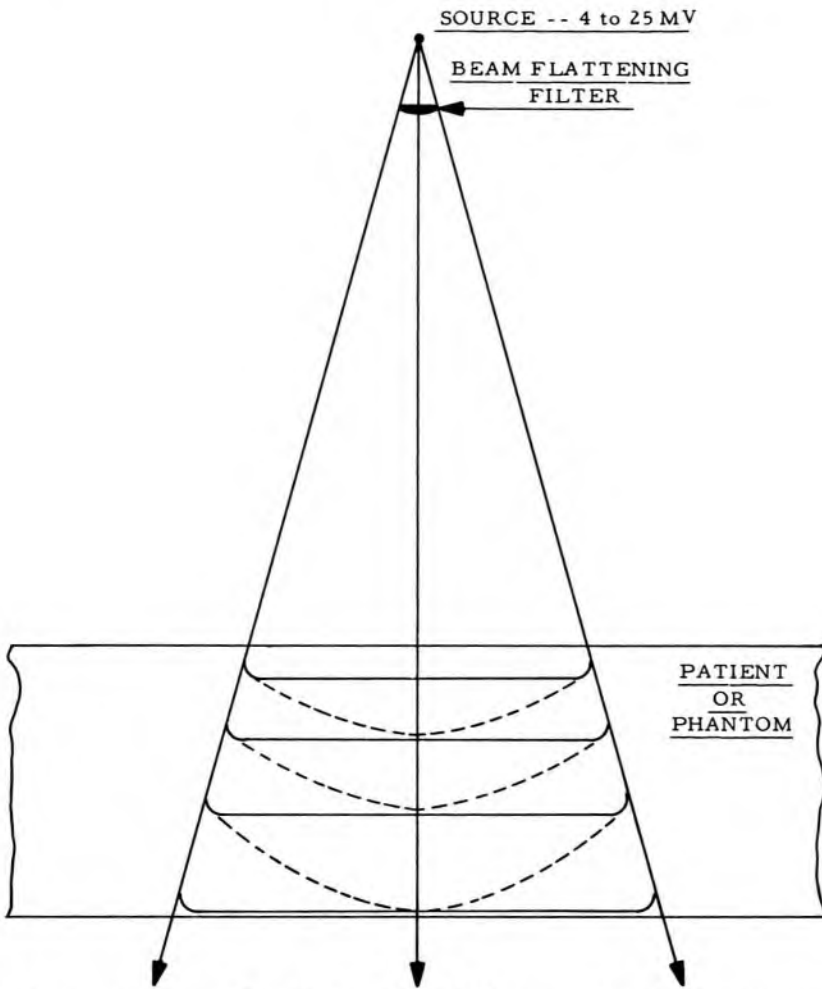


Fig. 16. Isodose curves for 4 to 25 MV x-ray beams—general shape.¹⁷ Dotted curves: isodose curves using x-ray beams as produced, without beam flattening filter. Solid curves (above dotted ones): isodose curves obtained with specially designed beam flattening filter in place.

are of most use in supervoltage therapy, when the use of bolus as previously described (p. 243) is impractical without sacrifice of skin sparing. As a substitute for bolus, wedge-shaped metal absorbers are placed closer to the source, at least 15 cm distant from the patient's skin. (A relatively small number of such wedge filters may be employed in practice to serve a large variety of treatment situations.)

An important additional use of wedge filters is the tailoring of summated isodose distributions. Figure 17 (Left) shows an isodose curve for a 5×5 cm cobalt-60 field, 80 cm SSD, using a 45° wedge filter. The wedge is seen to make the beam isodose lines oblique, but they remain relatively parallel to each other. The modified field is often called a "wedge field" because of its shape.

Such wedge fields are commonly used in two applications. The first is with obliquely incident beams to correct for the departure from

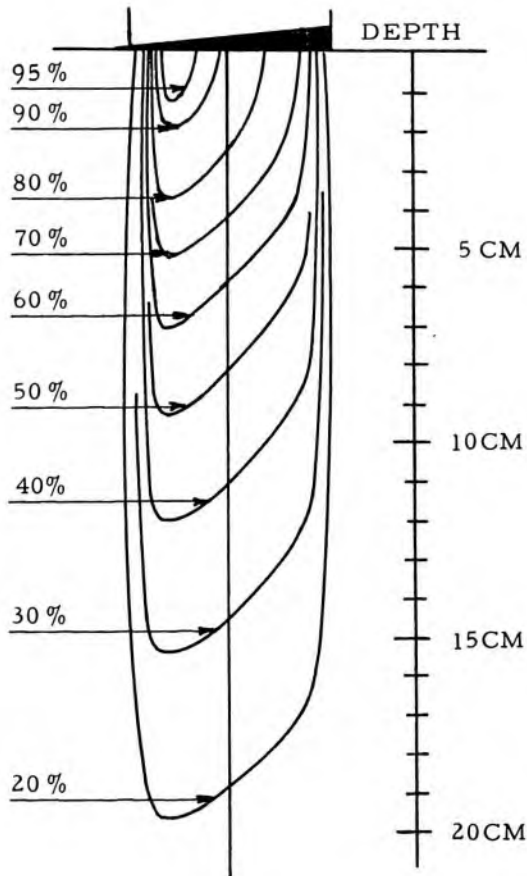
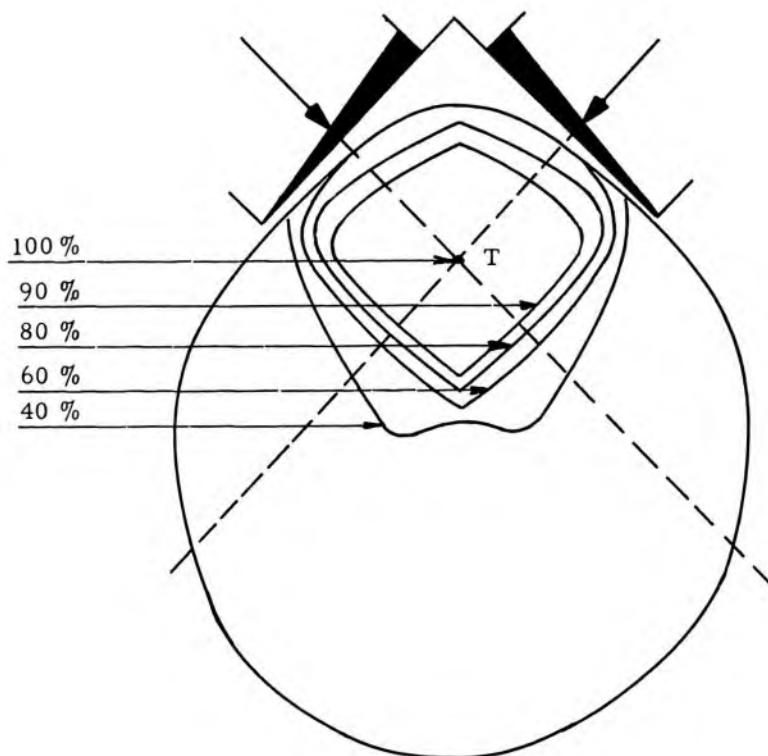


Fig. 17. Wedge filter principles — isodose charts. Left. For cobalt-60 field: wedge filter in place (5×5 cm, 80cm SSD). Below. Use of two wedge filters to deliver homogeneous dosage to a larynx carcinoma from two portals. (Redrawn from Johns, Morrison, and Watson.¹⁹)



perpendicular of the central ray. The filter metal then essentially substitutes its absorption for that of the missing bolus and restores the beam symmetry about the central ray.

A second, interesting use is to obtain homogeneous irradiation of a relatively superficial tumor site with a limited number of portals. Figure 17 (Right) shows the use of two wedge fields for cobalt-60 therapy of cancer of the larynx. Note the remarkably homogeneous dosage distribution in the treated area, virtually tailored to the anatomic area by the use of wedge filters.

A full discussion of the use of wedge filters is beyond the scope of this text, and the reader is referred to appropriate references for more information on the subject.

COMPENSATING FILTERS.²⁰ These are sometimes employed to assure relatively uniform dosage deep in the body to an extended target volume whose parts lie behind different thicknesses of tissue. Such an object could be a treated length of upper esophagus, which receives a very inhomogeneous dosage if one simply employs a long narrow single field. Figure 18 shows how the midline dose varies with and without a compensating filter. Note how a variable thickness metal absorber serves to compensate for anatomic differences in absorption, thereby assuring a relatively constant dosage throughout the full length of the tumor.

A relatively simple filter thus achieves a remarkable degree of dosage uniformity in a very difficult treatment situation. The inhomogeneity is reduced from 5 in 15 R per minute to something well below 1 in 12. In view of the unfortunately poor results in the treatment of tumors in this location, any reasonable change in the treatment procedure which could conceivably improve the result is most welcome.

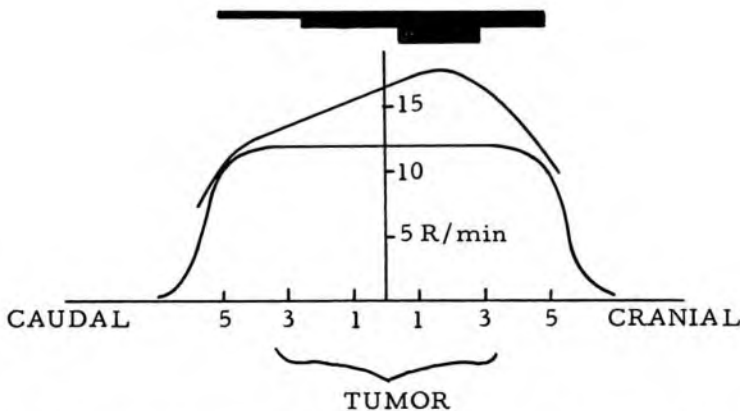


Fig. 18. Use of a compensating filter for esophagus treatment. Upper curve: no compensation filter. Lower curve: with compensation filter shown above curves.²⁰

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8

Physics of X-ray Diagnosis

As in all medical diagnosis, the object of an x-ray study is to obtain maximum useful information with minimum hazard to the patient. The medical knowledge and skill of the roentgenologist is one prerequisite to achieve this goal. An additional prerequisite is knowledge of relevant physical and engineering principles relating to three broad categories: production of the image; x-ray equipment, accessories, and films; and special techniques. This chapter attempts to provide a conceptual framework for such a background. It must be emphasized that rapid progress makes continued further study essential to keep up-to-date. More complex problems often require the cooperative efforts of radiologists, engineers, and medical physicists. In this collaboration the medical physicist may be especially helpful because in addition to his own scientific contribution he brings a knowledge of the terminology and philosophic approach of both other disciplines.

In this chapter we provide a brief systematic discussion of five major aspects of x-ray diagnostic physics. Fundamental concepts are stressed and unnecessary engineering details avoided. The sequence of topics follows:

1. The x-ray image, its production and quality.
2. Basic tools of x-ray diagnosis, fluoroscopy and radiography.
3. Electronic x-ray imaging systems.
4. X-ray films.
5. Special diagnostic techniques.

THE X-RAY IMAGE

Central to making the diagnosis is the x-ray image viewed by the roentgenologist. This image must of course contain the essential information upon which the decisions are made. However, this information must additionally be in readily usable form to facilitate pat-

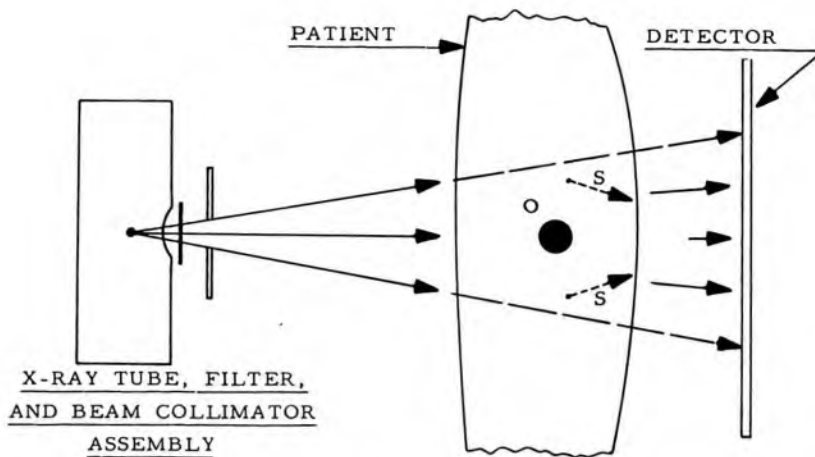


Fig. 1. Basic setup in a diagnostic study. X-rays traverse the patient and strike the detector. This can be a fluorescent screen, intensifying screen cassette, directly exposed film, or image intensifier. The final result is examined by eye as either a roentgenogram or luminescent image on a fluorescent screen, image tube phosphor, TV tube, or ciné projection screen. The final image of object *O* reflects differences in x-ray attenuation between it and surrounding tissues. Lengths of solid arrows represent directly transmitted intensities. In practice, contrast is greatly reduced by scatter (*S*, dotted arrows).

tern recognition—easily seen, with a minimum of distracting details, and viewable with minimum fatigue.

This section first discusses how the x-ray image is produced, and then the nature and important clinical aspects of image quality.

X-Ray Image Production

Figure 1 shows the basic exposure setup in any diagnostic study. X-rays from a relatively small source traverse the part of the patient which is of interest, and the emerging beam strikes a detector. This detector uses the fluorescent or photographic action of x-rays (or both) to derive a visible image that corresponds to the attenuation characteristics of tissues in the patient.

THE X-RAY BEAM IMAGE. The emerging x-ray beam reflects the normal and pathologic human anatomy of the patient and is inherently complex. Nevertheless, it is instructive to first consider the simplest possible case, that of a *single* object in a *uniform* absorber (Fig. 2). The object introduces an atypical transmission pattern or contrast into the transmitted beam. From our previous discussion (Chap. 3), it will be recalled this subject contrast results from a *difference* in transmission of the objects and its surroundings. Subject contrast depends directly on $(t\Delta\mu)$ (Fig. 2) where:

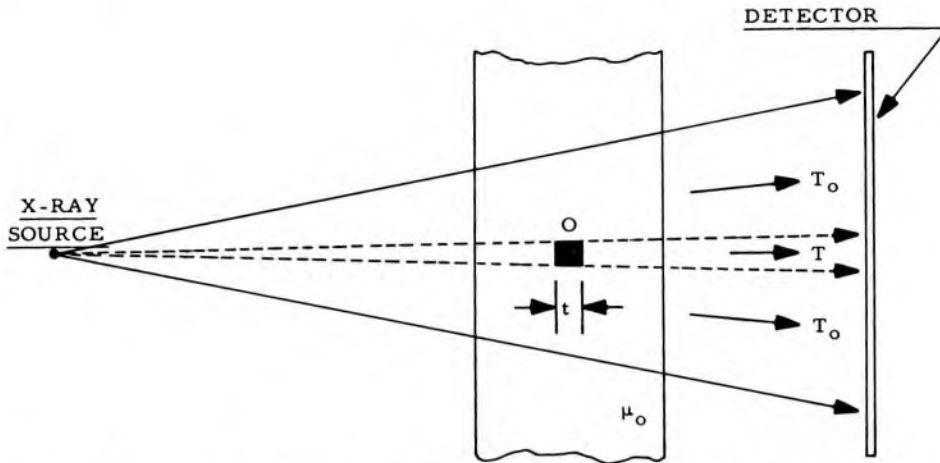


Fig. 2. Production of a simple x-ray image. A small object O of atypical attenuation coefficient μ , t cm thick, is shown in a uniform medium of attenuation coefficient μ_0 . Transmission, for narrow beams through medium generally: $T_0 = e^{-\mu_0 t}$; through object O : $T = e^{-\mu t}$. Contrast then depends on $T/T_0 = e^{-(\mu-\mu_0)t} = e^{-t\Delta\mu}$. Hence, contrast originates in $(t\Delta\mu)$. Note scatter always tends to reduce this contrast in practice.

t = object thickness in centimeters

$\Delta\mu$ = the difference in linear attenuation coefficient of the object and surrounding material. These μ values in turn depend on the density and atomic composition of these materials and the beam quality.

In practice scatter can greatly reduce the final contrast. In x-ray studies of 15 to 20 cm body parts with large portals, the scatter intensity at the film is between 2 and 4 times *greater* than the primary radiation. It is evident that grids are essential in such work; the general object is to reduce scatter to about 20 percent or less of the total x-ray beam intensity at the detector. Since a grid removes so much of the x-rays otherwise reaching the film, exposure mAs values are generally increased 3 to 6 times in switching to a grid technique if other factors remain the same. Generally, the amount of scatter increases slowly with kilovoltage, rapidly with irradiated volume (i.e., field area times part thickness).

THE DETECTOR SYSTEM. We have thus far spoken of the *x-ray* image, which is essentially information in the x-rays striking the detector. Detectors, however, can introduce major changes. Some of these are useful; obviously, providing a visible image is an indispensable contribution. In addition, film and television systems can yield contrast enhancement, highlighting details otherwise lost. (Simple fluoroscopic screens cannot do this.) However, we lose something in the process: resolution of detail is to some extent always sacrificed.

This is a crucial concern in certain studies, eliminating some otherwise preferable detectors from consideration.

Image Quality

There is ordinarily a great deal of laxity in discussing roentgen image quality. Inadequate definitions, such as "sharp," "crisp," and "washed out," are common. We shall now define and discuss the following, more specific terms: image contrast, blur, noise, resolution, and latitude.

BASIC DEFINITIONS OF TERMS. Consider the simplest situation: the image of a single rectangular object viewed in a film on a viewbox or as a fluorescent image (Fig. 3). There are two well-defined areas, each of relatively uniform brightness: the main image in the center and the surrounding areas. Generally, a transitional border exists around the main image, indicated as Δx wide in the figure. ΔB is the difference in brightness between that of the image ($B + \Delta B$) and surrounding areas (B).

We can now define the luminance contrast as $(\Delta B/B)$. This is the contrast information in the light reaching the eye, and is stressed here because *this* is what produces the visual stimulus. It of course

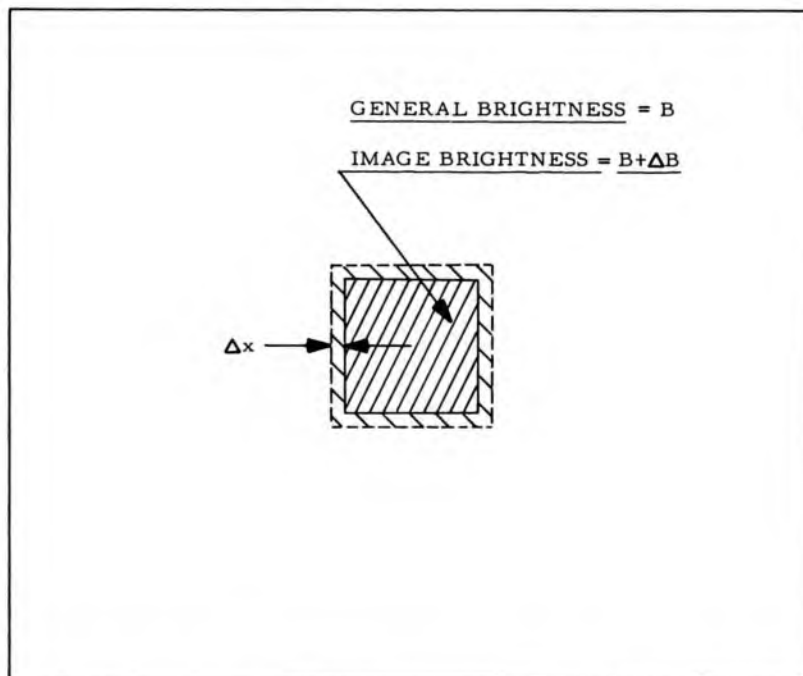


Fig. 3. Image of a single rectangular object as viewed on a viewbox or as a fluorescent image. Two well-defined areas are seen: the main image in the center and surrounding areas. A transitional "blur" area, Δx wide, is also generally present. We define luminance contrast as $(\Delta B/B)$ and the physical sharpness as $(1/\Delta X)$.

reflects the x-ray beam contrast ultimately, as modified by the photographic and viewing processes. Luminance contrast is usually fairly high in most roentgenography (10 to 200 percent). The human eye can respond under favorable conditions of object size and general illumination to as little as 1 or 2 percent and less.

(ΔX represents the width of a *blur* in the margin of the image. The smaller ΔX , the "sharper" the image, so one could define $(1/\Delta X)$ as the *physical sharpness*.)

The visual stimulus depends on *both* contrast and sharpness. Thus, for good viewing conditions, the visual stimulus is given by:

$$\text{V.S.} = k \frac{(\Delta B/B)}{\Delta X} \quad (8-1)$$

This assumes image contrast, size, and general brightness to be adequate for good visual cone vision. In practice, light glare always reduces the visual stimulus because, in effect, it increases the value of B at the retina without a corresponding increase in ΔB .

Image recognition is generally hampered by *noise*, which is of two general types: false images and mottle.

False images are commonly referred to as "artifacts." They originate from mechanical, chemical, and radiation causes, as well as intensifying screen defects. All artifacts *obscure* the desired image. Mottle is *granularity* of the image, arising from two facts. First, the detector is composed of individual crystals (silver bromide in films, calcium tungstate in screens, etc.); second, the x-ray beam is composed of individual photons. The relative significance of these two causes of mottle, and of noise in general, varies greatly with the application, so this subject is discussed below in more detail in connection with specific detectors.

Both blur and noise affect image *resolution*. This refers to the fidelity of reproduction in the final image of small details present in the original object. High resolution is needed to demonstrate fine details such as bone trabecular detail and fracture lines, punctate calcifications and fine soft tissue detail in the breast, and fine blood vessels in cardiovascular opacification. Fine details are of course lost if they are obscured by artifacts or smeared out by unsharpness or blur; also, low contrast images can be "lost" in graininess of films or scintillation of image intensifiers if mottle is excessive.

ACTUAL IMAGES—LATITUDE. *Latitude* is the ability of an examination to satisfactorily demonstrate objects of markedly different roentgen absorption on the same picture or roentgenogram. It is very important in clinical radiography, which almost always requires essentially survey information of many organs in a single view. A chest

film, for example, should show ribs, spine, hilar soft-tissue structures, lung fields and upper respiratory tract, and more. Similarly, an abdominal film should demonstrate the fetal skeleton, radiopaque materials in the kidney, vascular system or gastrointestinal tract, etc. In each study, a single view must visualize many objects having a great spread of x-ray attenuation, and all with useful contrast. This can be accomplished only by showing *all* desired parts with relatively *low* contrast, to encompass them all on a single film. It is usually done by using moderately high kilovoltage techniques.

Contrast and latitude are reciprocal quantities—one can be changed simply only at the expense of the other. Since most examinations require considerable latitude to encompass a wide variety of organs, higher kilovoltage technique is usually employed in roentgenology of the torso. Low kilovoltage, high contrast technique is usually reserved for those studies in which detection of relatively fine details is desired, usually in extremities. Contrast media are used in many situations to enhance contrast of selected parts, but this procedure cannot always be used.

From the above discussion, image quality essentially involves getting satisfactory contrast and latitude with adequate resolution. Intelligent technique design requires a knowledge of what determines these. We shall now consider the factors involved.

CONTRAST. Table 1 presents a breakdown of factors contributing to the final image contrast. Subject contrast and scatter have been discussed above, and the way the factors contribute to the contrast follows directly from the nature of x-ray attenuation as described in Chapter 4. The contrast enhancing action of some detectors is, however, less obvious.

The brightness of any fluorescent screen is roughly proportional to the x-ray intensity. For this reason, screens do not enhance contrast because a 1 percent increase in x-ray intensity yields only a 1 percent increase in screen brightness.

What about an image intensifier tube? As will be shown below, such tubes act essentially to derive a *brighter* image from an initially dim one produced in another part of the tube. All parts of the new image are brighter than the original one *by essentially the same factor*. Consequently, each image and its surrounding areas are also brightened to the same extent, and the contrast is unchanged. This is analogous to examining an overexposed film on an adjustable high intensity viewbox. One turns up the brightness to obtain optimum retinal illumination, and of course the luminances of all parts are thereby increased by a constant factor. In each case, however, ΔB and

TABLE 1. FACTORS DETERMINING CONTRAST OF THE IMAGE

A. Subject Contrast

1. The object and surrounding material
 - (A) their chemical compositions (Z)
 - (B) their densities (ρ)
 - (C) the object thickness (x)
2. The x-ray beam spectrum
 - (A) operating tube pkV; wave form—pulsating vs. constant or 3-phase voltage
 - (B) filtration

B. X-ray Beam Scatter

1. Total tissue volume irradiated: field size and patient thickness
2. Beam quality: effectiveness of grids is reduced with harder beams.

C. X-ray Detector Used

1. Some do not affect contrast: fluoroscopic screen and directly viewed image intensifier tubes
2. Others do affect contrast:
 - (A) X-ray films, used with screens. Contrast increases of 2 to 3.5 times. True for cineradiography as well.
 - (B) X-ray films, directly exposed. Contrast factor of roughly 2.3 times the film density. (Up to 7 times, for very dark films of density 3!)
 - (C) Any TV systems: substantial contrast gain possible, as with film.

B increase by the same constant factor; $(\Delta B/B)$ stays the same, so contrast is unchanged.

Both x-ray films and television viewing systems can enhance contrast. Films do this by virtue of both their characteristic response curves and their being viewed by *transmitted* light (see below). Television viewing systems consist of three essential elements: a television pickup tube, amplifier system, and cathode ray tube display. Any of these can by design be made nonlinear in response and yield contrast enhancement. In practice, television system fluoroscopy yields much higher contrast images than direct-viewing image intensifier fluoroscopy. This is readily observed by direct comparison.

Sometimes roentgenograms contain information not readily gleaned by ordinary viewing methods. This can be either because the latitude has not been sufficiently great (too high contrast) or a given part has too *low* contrast. Special methods exist for deriving more suitable roentgenograms from the original, including subtraction tech-

niques and photographic or electronic "dodging" (see below). It should be emphasized that no "dodging" methods can provide information not originally present; in fact, some information is always lost in any derived image.

RESOLUTION IN GENERAL. Detail resolution must be adequate for the task, or essential diagnostic information may be lost. The basic problem has been emphasized by development of newer electronic imaging systems which are increasingly replacing ordinary fluoroscopy and radiography equipment (see below). The proper evaluation of these systems involves considerations of not only resolution, but contrast and patient dosage as well. Visual physiology, perception, and pattern recognition are also involved in any system evaluation.

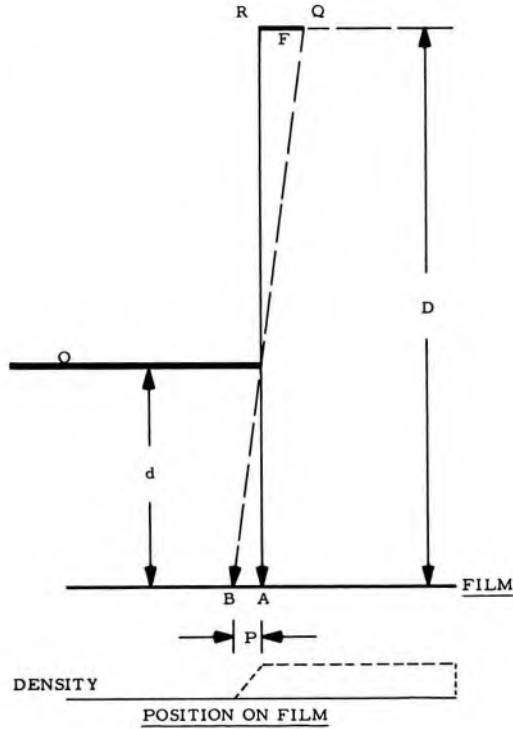
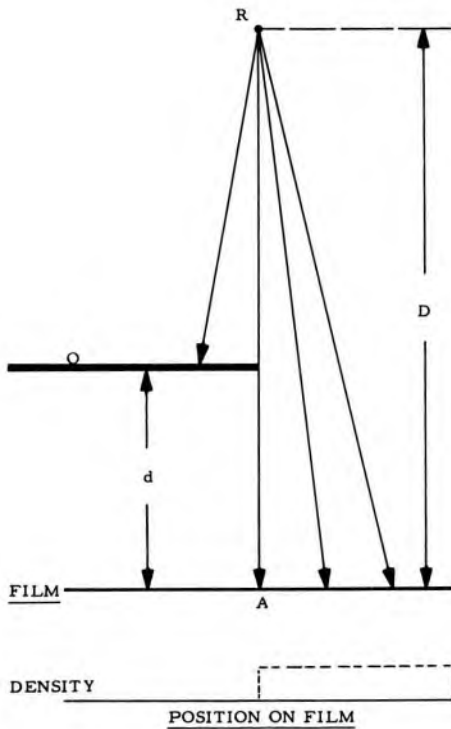
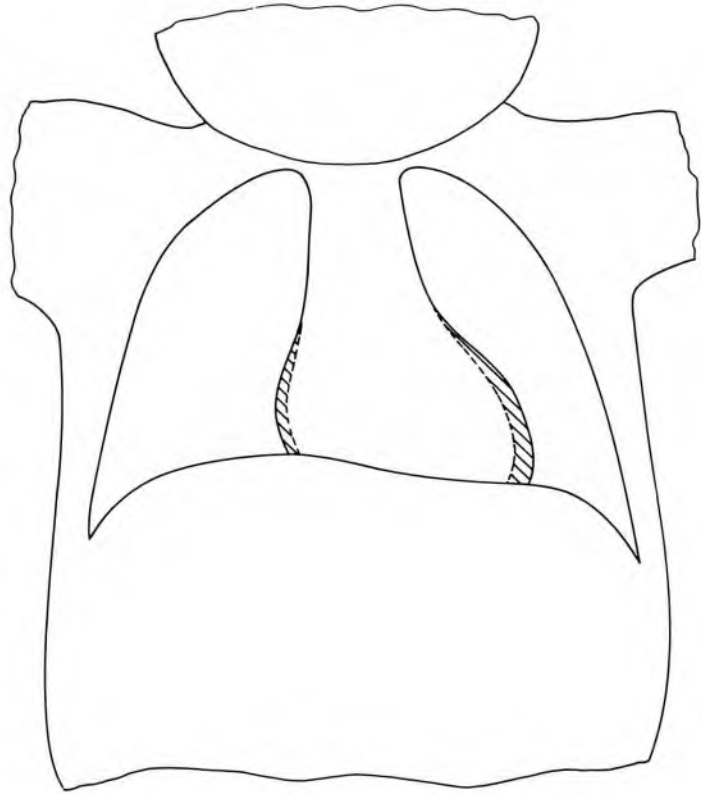
Our discussion is necessarily short. It will be confined to a brief explanation of image blur and noise, the two quantities that limit detail resolution. The former can arise in both the x-ray exposure (motion and penumbra blur) and in the detecting and processing system (detector system blur). Noise can be caused by artifacts and mottle.

MOTION BLUR. Motion blur is familiar to all photographers, amateur and professional alike. Difficulties arise in x-ray work in both single exposures and sequential studies.

Figure 4(Top) illustrates motion blurring in chest radiography (single exposure). Shown is a tracing of an infant's chest radiograph, indicating the heart and lungs. The extreme locations of the heart margin are indicated: diastole by dotted, systole by solid lines. The shaded areas represent the range of normal motion. A substantial blur is hence possible in the roentgenogram if exposure times are excessive. The blur magnitude, in general, depends on the nature of the movement and the exposure time. It is evident that the shorter the exposure time, the better when there is a cyclical effect like heartbeat or pulsatile vessel motion. Higher kilovoltage techniques with moderately great tube currents permit use of quite low exposure times (1/120 second and less). In some other situations, requiring longer times and patient cooperation, a skillful x-ray technologist is indispensable to a successful study.

Another blur problem arises when the detector is too slow to respond to the events being studied. For example, image intensifiers used with earlier design vidicon television pickup tubes resulted in "smeared" images in study of rapidly developing phenomena (cardiovascular and gastrointestinal studies, etc.). More recent vidicon television units are somewhat faster in response, but the more expensive image orthicon is still preferred by some radiologists for this reason.

Fig. 4. Image blur, due to motion and penumbra. Right. Motion blur. Tracing from infant chest exposure taken at 0.1 second, showing possible motion of ventricle walls during systole (shaded area). Below, left. Point source. No penumbra is observed since the x-ray intensity falls abruptly from maximum to zero at A. Below, right. Actual source. X-rays originating at R all irradiate the film to the right of A, none to the left. However, those originating at any other point on RQ produce irradiation to the left of A and as far left as B. For a rectangular source, a uniform intensity variation occurs from A to B. This is P, the penumbra, whose width is given by the expression, $P = F \frac{d}{D-d}$.



PENUMBRA BLUR. Penumbra blur is caused by the use of x-ray sources of finite size. In Figure 4(Bottom, left) the x-rays originate at a single point R. Note there is a sharp image because rays directed to the right of A all strike the film, while those directed to the left of A are absorbed by the opaque object O. The density on the film, for this idealized case, rises abruptly from zero to the left of A to maximum value at A and beyond. Figure 4(Bottom, right) shows the actual situation, with a focal spot of finite size, RQ. As before, rays originating at R can irradiate the film to the right of A, but not to the left. A ray originating at Q, however, can irradiate the film as far left as B. Also, photons originating *between* R and Q can irradiate corresponding points *between* A and B. All photons from the source can irradiate parts of the film to the right of A; *none* can irradiate parts to the left of B. The net result is an unexposed film to the left of B, fully exposed to the right of A, with continuously increasing exposure between B and A.

The region AB is called the *penumbra*, designated P. Since triangles ORQ and OAB are similar, one can easily derive the simplified expression for P:

$$P = F \left(\frac{d}{D-d} \right) \quad (8-2)$$

As an example, consider a certain chest technique in which $D = 100$ cm, $d = 15$ cm, and $F = 2$ mm. Then, $P = 0.35$ mm. Note that this penumbra blur is quite substantial compared with 0.1 mm, which is readily resolved by the unaided eye under good film viewing conditions. This significant penumbra is evidently quite tolerable for most chest roentgenography.

DETECTOR BLUR. Both motion and penumbra blur arise during the formation of the x-ray image and are therefore independent of the particular detector used to render this image visible. Such detectors, however, also introduce their own blur into the final viewed image. The amount of blur varies greatly with detector type and specific design, and this entire subject is currently under investigation by physicists and engineers. Detector blur is of great practical importance, so we shall consider the characteristics of the most commonly used detectors. The figures given below present a somewhat oversimplified representation of published line-spread and modulation transfer data. Nevertheless, they provide a useful gross indication of the relative ability to resolve detail of the detectors considered.

- a. *Fluoroscopic screens.*¹ These are relatively poor in resolution compared with other detectors; a blur of the order of 1 mm

width is introduced. Mottle is normally not noticed because the screen blurs it out.

- b. *Image intensifiers.*¹ These are of course under continuous development. Botden's data indicate a typical blur of the order of 0.5 to 0.7 mm width.
- c. *Intensifying screen-film combinations.*¹ These range from about 0.25 mm blur width for detail screens through 0.35 mm for medium and 0.5 mm for high speed screens. This of course applies to good screen contact; blur is much worse with poor contact.
- d. *Direct exposures of films.*² Blur width for directly exposed film is markedly lower than for screen-film combinations or image intensifiers, generally well below 0.1 mm. This is because the more complicated systems start with a fluorescent light image which then in turn acts on a film or photoelectric surface. The fluorescent light is inevitably diffused somewhat in the process, greatly reducing the final image sharpness.

NOISE. This term refers to both artifacts, or false images, and mottle from statistical effects.

Artifacts arise primarily in radiography, from improper film storage, handling, and processing. Dirty or damaged intensifying screens can also contribute major difficulties. Following manufacturers' recommendations can reduce such artifact problems to minor proportions.

Mottle refers to a spotty nonuniformity of the image even when the measured x-ray intensity is constant across the beam. As mentioned previously, it arises from the fact that detectors all consist of individual crystals and the x-ray beam of individual photons. It is interesting to consider its practical significance with different detectors.

Mottle is not troublesome in either direct fluoroscopy or screen technique radiography because, as shown above, all images are so indistinct that fine mottle is simply blurred out (as a skin blemish is blurred out in a soft-focus portrait photograph). However, in direct film exposures, significant "graininess" or mottle is observed when fast no-screen film is used for mammography. Slower industrial films exhibit less "graininess." This results from two factors. First, the required longer exposure results in delivery of more photons to produce the image; this reduces *photon mottle*. Second, the individual silver halide crystals are smaller, reducing *crystal mottle*.

Mottle is perhaps most troublesome in high gain intensifier systems. In the search for low patient dosage, radiologists have on occasion worked at very low x-ray output, particularly in ciné-radiography

applications. Unfortunately, the lower the x-ray tube milliamperage employed for a given kilovoltage, the fewer photons are received by each ciné frame. It is very instructive to compare ciné records taken at high and low mA values; the low mA ciné records are reminiscent of the "snowy" television image of a distant (low intensity) transmitter station as compared with the more distinct image of the same network program from a nearer station.

Clinical Aspects of Image Quality

Technical defects of roentgen images can spoil resolution of image detail, which can be very important in studies such as some bone examinations and mammography, in which detail is critical. Relatively coarse detail resolution is adequate, however, for the great majority of roentgenographic studies, permitting the general use of screen techniques.

Tuddenham³ has noted that capable radiologists miss diagnoses primarily for reasons other than lack of detail resolution. These reasons fall into two general categories:

1. *Low contrast at the eye.* This can result from an inherent low contrast object; obscuring details may present excessive contrast; contrast may be lost in an underexposed part of the film where film response is poor; and glare may reduce contrast at the retina. The image is essentially not *perceived*.
2. *Too much detail.* The brain can be overwhelmed by too much information irrelevant to the clues being sought. This complicates *pattern recognition*.

Low contrast can sometimes be helped by such procedures as exposing films to a greater density and using a minifying lens to increase the visual stimulus. However, the most general problem involves obscuring and confusing detail. Multiple views and ingenious exposure procedures are used to increase information and accuracy of diagnosis. These are briefly considered below in the section, "Special Diagnostic Techniques."

BASIC TOOLS OF X-RAY DIAGNOSIS

During the past seventy years much effort has been expended to develop x-ray equipment, accessories, materials, and procedures. We shall briefly discuss some of the more important developments in basic fluoroscopy and radiography in this section. Electronic systems, films, and special techniques are considered in later, separate sections.

Fluoroscopy

Historically, improper equipment and procedures have caused more injuries to both patients and operators in fluoroscopy than in any other type of x-ray work. Considerable skill and knowledge, both technical and medical, are required to obtain useful information. Good fluoroscopy, therefore, requires good equipment operated by a thoroughly trained and experienced physician.

EQUIPMENT. Consider the equipment employed, starting with the x-ray tube (Fig. 5; see also Fig. 3, Chap. 3). Most modern tubes are shockproof and rayproof. Tubes must be at least 12 inches from the table, to protect the patient's skin (see Chap. 14). Larger tubes have adequate ratings for both reasonably intensive fluoroscopy and spot-filming, barring abuse, and air blowers can further increase fluoroscopic ratings. Older nonshockproof fixed anode tubes do *not* meet modern radiation safety codes and should be replaced with newer tubes that do. The x-ray beam is controlled after leaving the tube by adjustable lead shutters which determine the field size of the beam. Both the lead shutters and the rest of the collimating assembly must move with the fluorescent screen assembly to prevent accidental operator exposure to the direct x-ray beam.

The fluorescent screen assembly includes the fluorescent screen with its lead glass shield. The screens use zinc sulfide crystals (with cadmium activator) mounted on a plastic base. This particular phosphor material is used because its light intensity peaks at 5,300 Å, very near the peak sensitivity of the retina in dim illumination. (When the intensity of illumination falls below the threshold of the cones, they virtually cease to function, and visual acuity then depends solely on the rods.) Screens may deteriorate with age and illumination (particularly older types) and should be periodically checked by comparing their image with that of a new screen held in the beam. It is good practice to keep screens covered when not in use; this incidentally also protects the rather soft lead glass shield from scratching.

The lead glass should be of adequate equivalent lead thickness for the kV employed. For example, a lead glass barrier adequate at 100 pkV requires addition of 0.5 mm lead equivalent lead glass to attenuate sufficiently at 150 pkV. It must be stressed that a great operator hazard can result with inadequate shielding of both direct and scattered x-rays. (See Chapter 14.)

Except for orthopedic work, fluoroscopy usually involves irradiation of thick parts of the body, so a grid is generally needed to control scatter fog. This is mounted below the cassette holder. Motor-driven or "reciprocating" grids are often used to avoid grid lines on spot or

flash radiographs. Fixed "fine-line" modern grids are often used nowadays.

Mechanical arrangements are provided to permit taking exposures on different areas of the same film. Cassette "tunnel" arrangements are mounted below the screen. A cassette is initially stored in a shielded location of this tunnel during fluoroscopy. When the radiologist decides radiographs are desirable, he moves the cassette into the field and takes carefully positioned views. The cassette is moved in and out of the field by specially designed hand- or motor-operated "cassette changers."

Preceding and following fluoroscopy, it is often desirable to also radiograph the general area fluoroscoped in the conventional manner. For this purpose one uses a second overhead tube connected to the same machine. This necessitates an arrangement for moving a cassette and grid unit ("bucky tray") lengthwise beneath the table. To permit inserting cassettes, this unit must be accessible from the side, so a fairly wide slot, called the "bucky slot," exists near the table top (Fig. 5). In fluoroscopy this constitutes a defect in the fixed shielding against patient scatter. Some machines are therefore provided with a sheet metal shield for fluoroscopy, which retracts when the bucky tray is placed beneath the patient.

PROCEDURES IN GENERAL. During fluoroscopy the exposure time, field size, and x-ray intensity should all be carefully controlled. The writer has been assured by many skilled radiologists that one perceives all he is likely to observe in a relatively few seconds, and further viewing of the same local area involves useless irradiation of the patient. A sequence of multiple short exposures is therefore considered to be the most useful. This is evidently because fluoroscopy is best for viewing *motion*, which the eye and brain instinctively grasp almost instantly. In addition, fluorescent screens exhibit "after image": the image persists briefly after the x-rays are turned off. *Detail search* requires good detail resolution and greater viewing time, both better provided by radiography than fluoroscopy.⁴ Excessive field size is potentially harmful to the patient. In addition, it increases scatter, which both reduces useful information and increases scattered radiation to the operator. Finally, the x-ray intensity must be kept reasonable. Even under the best circumstances, the patient's skin receives something of the order of 5 R per minute of fluoroscopy. If the exposure rate and duration are excessive, therapeutic dosage levels can be given to the skin!

Two situations exist in which the inexperienced operator is most tempted to raise kV or mA levels excessively. The first involves fluoroscopy without visual accommodation. As mentioned above, fluoros-

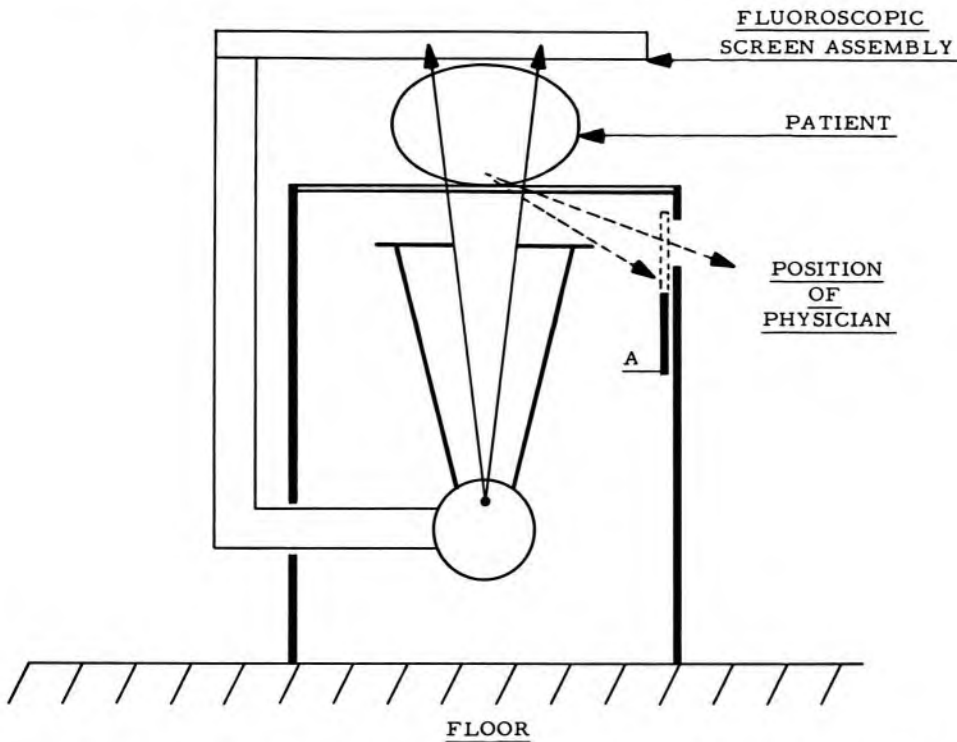


Fig. 5. Scatter to physician from patient, through bucky slot. A slot is required in the table to permit inserting cassettes in the "bucky tray" (not shown) from the side. Note how backscatter from the patient can reach the physician unattenuated by the table side structure. Some machines are provided with a shield (A) which covers this slot during fluoroscopy (dotted lines). It retracts automatically to position shown when the bucky tray is moved in place to perform radiography with the x-ray tube above the table.

copic images are very dim, and normally only rod or dark vision is employed. The retinal rods are bleached into insensitivity at ordinary illuminations, such as those employed in direct patient examination, reading, and viewing radiographs. The eye, therefore, requires at least 15 to 20 minutes during which the rods are protected, permitting restoration of rhodopsin levels. Without this preparation the eye is insensitive to dim light, and a fluoroscopist is tempted to raise the screen brightness unduly. Of course, this is at the cost of gross over-irradiation of the patient.

Fortunately, the use of suitable x-ray red goggles permits the physician to do work at nearly normal illumination intensities while still sparing the rods. Figure 6 shows how this is possible.⁵ The curves show that retinal rod vision (and hence bleaching of rhodopsin or visual purple of the rods) occurs most effectively at 510 millimicrons wavelength (green light) but is minimal at 620 millimicrons wavelength and higher red light. However, such red light is readily seen by the cones. Special red goggles remove light of wavelengths

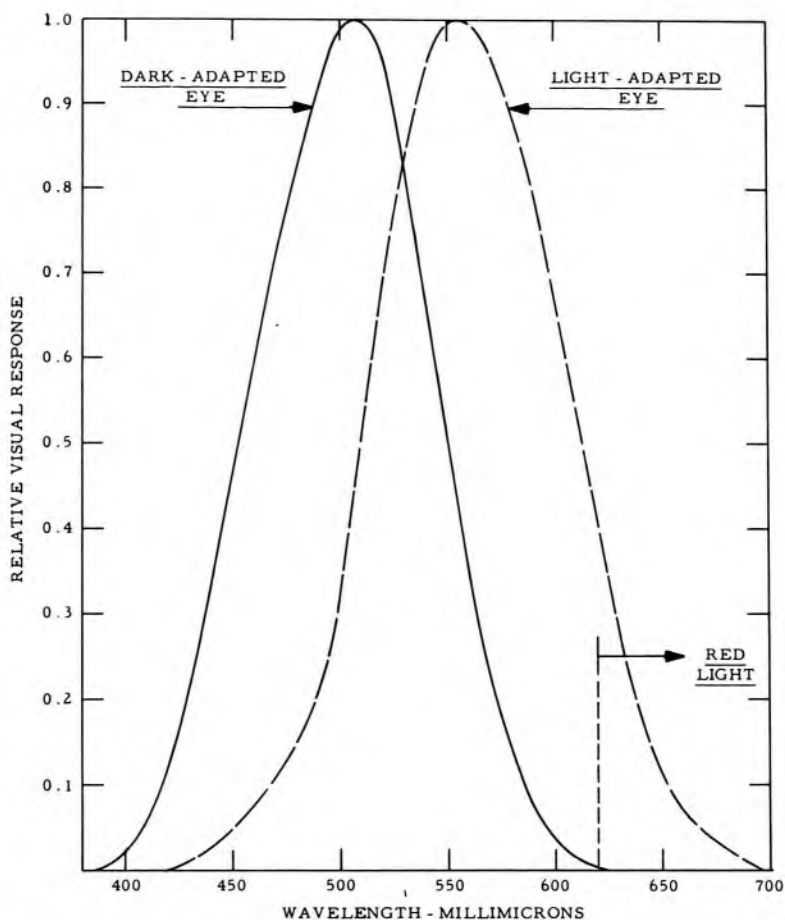


Fig. 6. Response curves of dark- (solid) and light-adapted eye (dashed) to various wavelengths of light. Note the rods (dark-adapted) are not sensitive to red light, 630 millimicrons and greater in wavelength, but the cones are. Hence red goggles protect the rods from shorter wavelengths while transmitting red light. Since the cones *do* respond to red light, the radiologist can carry out many functions while becoming dark-adapted. (Redrawn from Hecht and Williams.⁵)

620 millimicrons and less, so the radiologist wearing them sees objects by means of his retinal cones while his rods are spared. Such goggles are not perfect light filters, but in practice they are adequately effective.

A few comments should be made about the protection of personnel. All persons participating in fluoroscopy should wear lead aprons for protection against scattered radiation, and only the physician and other necessary personnel should be in the room. The physician's apron must be a heavy type (0.5 mm lead equivalent) and preferably cover the shoulder area, a necessary measure because he works close to the patient. Others may wear lighter aprons (0.25 mm lead equivalent). The radiologist also requires lead gloves, as well as special lead hand shields in some catheterization procedures. Normally, hands should be kept out of the direct beam. On occasion when

essential to the examination, they should be on the *screen side* of the patient, away from the tube. This uses to advantage the fact that the patient's torso absorbs at least 90 percent of incident x-rays. Finally, lead-rubber flaps are mounted on fluoroscopic screen assemblies to further protect the fluoroscopist from scattered radiation. Further information on protection in fluoroscopy is given in Chapter 14.

PROCEDURES TO INCREASE INFORMATION. Several procedures are routinely employed in fluoroscopy and spot filming to insure greater yield of diagnostic information. Contrast media are generally required for fluoroscopy because soft tissues of inherently low contrast are examined. Media used include barium sulfate in water suspension and aqueous solutions of iodinated hydrocarbons. Iodinated oils, used both directly and as emulsions, are also employed for certain studies. Catheters for selective visualization of regional circulation are made of plastics containing radiopaque "fillers," to facilitate their fluoroscopic localization.

Gas distension is useful in revealing details of mucosal patterns in studies of the stomach, rectum, and colon. In the upper gastrointestinal tract the stomach is first partially filled. Then, air is added, usually by swallowing. In the lower tract the ordinary barium sulfate study is first performed and the patient allowed time to empty the organ of interest. After most material has left, some often remains within the mucosal folds and on polyp surfaces. Gas is then inserted appropriately to distend the lumen, so residual barium salt in mucosal folds becomes visible as discrete absorbing objects.

A simple but important procedure is abdominal compression during fluoroscopy. Both compression cones and hands are used to squeeze out excess barium sulfate from and suitably position visceral organs. This assures optimum mucosal coating and organ positioning.

A final comment about preparation of patients for fluoroscopy. The finest radiologist cannot successfully detect a gastrointestinal lesion in the presence of excessive obstructing material, such as fresh food and fluid, feces, and gas. Changes sought are generally fairly subtle, and proper preparation of the patient is essential to avoid repeat examinations with consequent additional patient irradiation.

Radiography

Potentially radiography can yield both excellent resolution and a permanent record with much less patient dosage than fluoroscopy. For this reason the study of fixed body parts is best done radiographically. Spot film and other radiographs, furthermore, add greatly to the

usefulness of fluoroscopic examinations. In many special studies the progress is desired of a bolus of contrast medium following injection by catheter into a selected artery; rapid sequential radiography is carried out for this work. We shall first discuss equipment and tools of general radiographic work and then add a few comments about sequential radiography.

EQUIPMENT. A major equipment consideration in radiography, as in fluoroscopy, is radiation safety of the patient, involving both beam filtration and field size control.

For all but very low kilovoltage work an aluminum filter 2 to 3 mm thick is required.⁶ In very low kilovoltage work like mammography, however, the filter is generally omitted.

The primary beam must be well controlled by a cone or adjustable diaphragm at the tube. Cones use diaphragms to restrict the beam, with sheet metal conical attachments to indicate the resulting beam limits. Two warnings concerning cones are appropriate. Many old cones are defective in both design and construction, and cones with ineffectual *aluminum* diaphragms are occasionally seen. Others have diaphragms bearing no relationship to the fields indicated by the attached metal assemblies. It is good practice to expose a film to check the actual field. Some cones are initially good but become damaged in use (dented, tilted obliquely, etc.). Another consideration must also be borne in mind: even the most accurate cone is useful only at a *single* distance. A cone yielding a large angle beam is appropriate for some close distance radiography but irradiates an excessively large patient area at greater distances (Fig. 7). In practice, a good rule is to be sure always that the cone diaphragm prevents exposure of at least 3 corners of the film; this assures adequate collimation.

With the contemporary emphasis on reduction of patient gonad dosage, adjustable diaphragm units have been developed.⁷ These are basically similar in principle to those used in radiation therapy. They have a built-in light and mirror system; the light source is arranged so that the indicated light field on the patient closely matches the x-ray field. These devices remove collimator guesswork. Although some problems arose initially relating to mounting on tube assemblies, they are reasonably well solved in newer units. The mirror adds to the total filtration; this can be troublesome in mammography although it is not normally a serious problem. Technicians often complain routine work is slower with adjustable diaphragm units. However, it is likely these are not basic limitations, and this type of unit will be increasingly used in the future.

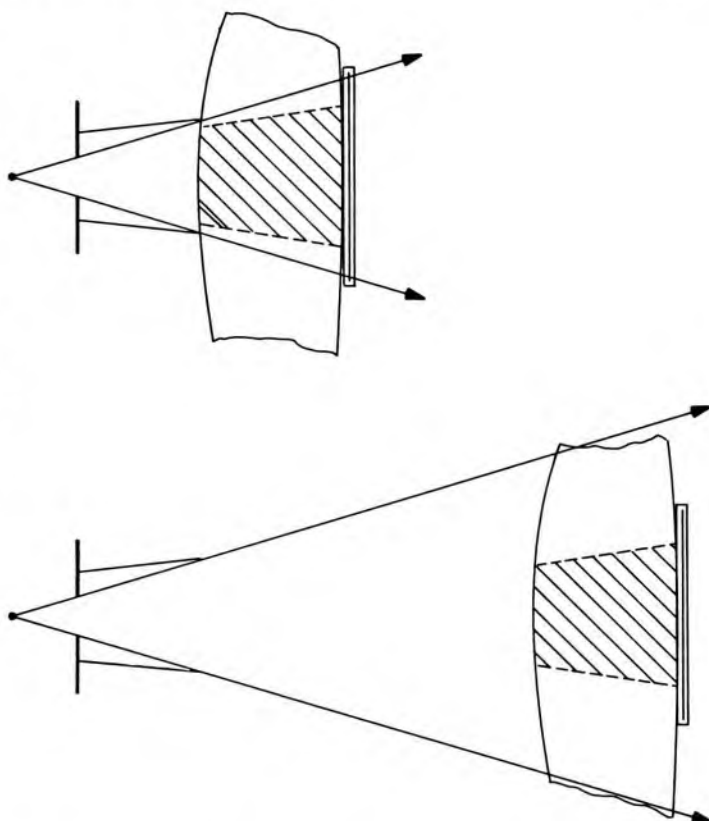


Fig. 7. Cone must be used at proper distance. Top. Proper distance. Proper overlap of volume of diagnostic interest (shaded area). Bottom. Too far away. Unnecessarily large area of patient irradiated.

In many examinations patient gonad shields are used. These usually consist of sheets of lead rubber which is a fabric of rubber containing lead oxide to absorb x-rays, and is also used in aprons and gloves. The lead rubber sheets are either draped or tied over the areas to be shielded. They protect primarily against the primary beam, the major gonadal radiation hazard in radiography.

GRIDS. Grids are required to prevent excessive image degradation from scatter. Their basic principle has been previously described (Chap. 3), but we shall now discuss their construction and application in somewhat more detail.⁸

A grid consists of an assembly of lead strips about 0.05 mm thick arranged in a "venetian blind" pattern (Fig. 8; see also Fig. 5, Chap. 3). They are held roughly 0.33 mm apart by spacers of special paper or plastic (in more expensive units, aluminum is used); the overall assembly usually has protective covers, and the total thickness ranges from 3 to 10 mm, depending on design. The slats may be inclined increasingly, to match the beam inclination, as one departs from the central ray. Such a grid, properly positioned, casts minimum shadows of the lead slats (called "grid lines"). The appropriate distance for minimum grid lines at the sides is called the "grid radius," and such a grid is called a "focused grid." When some marginal grid lines are permissible or small angular fields are

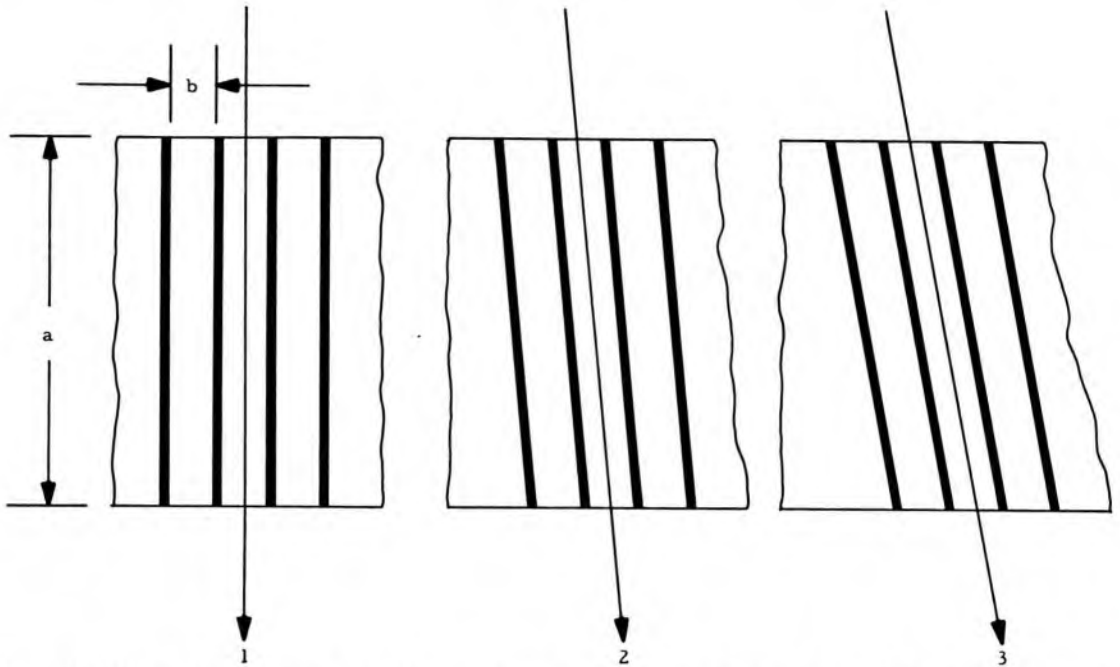


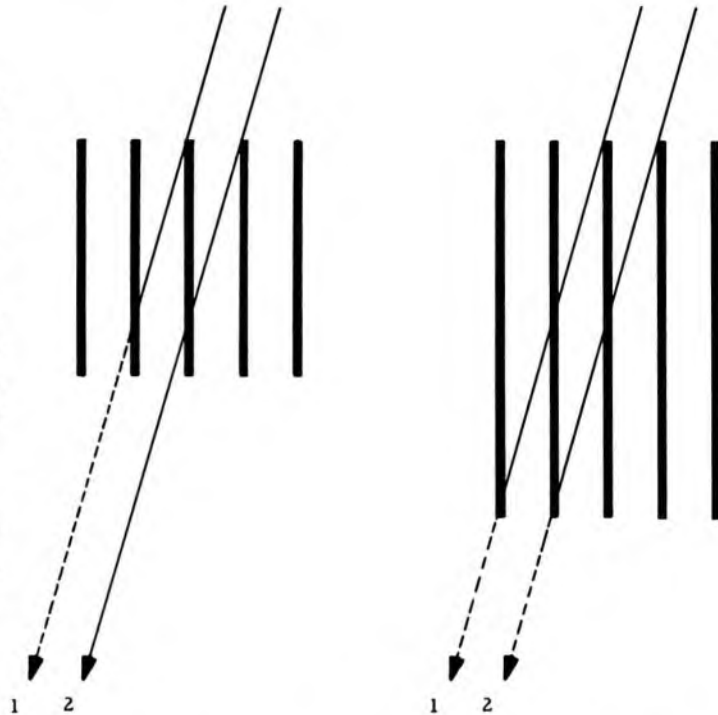
Fig. 8. Sketch of basic grid construction, to scale—8:1 grid. Grid ratio is (a/b) . Typical values are 5, 8, 12, and 18:1. Note increasing inclination of slats to side in focused grid: 1 is central ray, 2 is intermediate, 3 is edge area of grid. Perfect alignment of the edge slats with the beam occurs only at a certain distance from the target. This is called the “grid radius.”

used, simple grids without this elegant construction, in which the slats are all parallel to each other, may be used (“parallel” or “un-focused” grids). The ratio of the width of grid slats to their separation is called the “grid ratio” (a/b in Fig. 8).

The main reason for going to higher grid ratios is to assure proper cleanup at high kV operation (Fig. 9). Unfortunately, high grid ratios also increase the incidence of grid lines from improper centering, angulation, or distance from the tube. Also used at high kV's are “crossed grids.” These are essentially a pair of ordinary grids used together with their slats *perpendicular* to each other. Referring to Figure 3, Chapter 3, while one grid absorbs rays like #3 in the drawing, the other traps those like #5, which the first transmits. Actually, two crossed 8:1 grids are slightly more effective than a single 16:1 grid.

In practice, the choice of grid requires compromise among most effective cleanup, required image resolution and contrast, and dependability of the procedure with available personnel. Generally, 5:1 grids are most versatile, 8:1 yield better quality for general use; and 12:1 or 16:1 or crossed grids are used with higher kV techniques, preferably at longer TFD values to minimize grid lines.

Fig. 9. Why higher grid ratio improves scatter cleanup at higher pkV. Shown are sections of 5:1 and 8:1 grids. Left. 5:1 grid. 1. Scattered ray for 80 pkV operation, inclined 16° —readily attenuated by single lead strip. 2. Same ray at 125 pkV—only *partially* attenuated because of greater penetration in lead. Right. 8:1 grid. Note 16° ray must penetrate two lead strips to emerge from grid. Hence, both rays (1 and 2) are attenuated.



We have not yet discussed moving or bucky grids. Any fixed grid casts shadows which appear as narrow lines on the film. These were often troublesome in the past, but modern “fine-line” grids are often usable without motion for many purposes. Most radiologists, however, still prefer “bucky” devices for many studies. These move the grid during the x-ray exposure, thereby blurring the grid lines so they are not disturbing to the roentgenologist. The use of bucky grids requires following certain rules, to prevent production of grid lines by occurrence of part of the exposure with the grid either not moving or moving at just the right speed for “synchronism” with x-ray intensity fluctuations. Such rules are usually applied without great difficulty by a well-trained technologist and are not a topic of concern in this discussion.

INTENSIFYING SCREENS. The remaining item in the radiographic process is the cassette and film. This is a remarkably efficient combination, permitting radiography with reasonable exposure time and patient dose. (See Figure 6 in Chapter 3). Films constitute a rather large topic, reserved for a separate section below, so we shall confine the present discussion to intensifying screens. These consist of rather finely ground phosphor material (fluorescent crystals) incorporated in a plastic binder and supported on a cardboard or plastic sheet.⁹

Phosphors most commonly used are calcium tungstate (slow and medium speed) and barium lead sulfate (high speed). Speed

and resolving power of the screen depend on particle size and thickness of the phosphor layer. Typical values of the two quantities for medium speed screens are 5 and 100 microns, respectively. Contrary to many published assertions, crystal size is *not* the main limitation to film-screen resolution. It is rather light divergence in traveling from the screen to the film emulsion, particularly to that remote from the screen. Regardless of the specific reason, however, fast screens generally yield roughly *twice* the blur of medium speed screens, slow screens about *half*.

Ter-Pogossian has designed an experimental screen using thallium-activated potassium iodide (KI(Tl)) as a phosphor. This is five times faster than medium speed screens, with negligibly reduced resolution. The improvement is derived roughly equally from two sources: the higher mass absorption coefficient of iodine than that of both tungsten or lead for diagnostic x-rays and the higher light output per x-ray photon absorbed in KI(Tl). A theoretical speed increase of ten times is possible. If such screens prove to be practical from point of view of cost, ruggedness, and stability of response, they can contribute significantly to reduction of patient dose in radiography.

MANUAL EXPOSURE TIMING. To assure proper image quality, a film must be exposed for a time appropriate to the exposure factors and the patient size. Both manual and automatically preset timers are used to accomplish this.

Synchronous and electronic manual timers are employed. The former are essentially synchronous motors which open sensitive switches to terminate the exposure. Good synchronous timers are reasonably accurate down to nearly 0.1 second exposures. Below 0.2 seconds, however, the pulsating nature of the magnetic forces involved in the timer and contactor switch (since ac is used) limits the reproducibility of the switching action, at both the start and end of the exposure. For times 0.1 second and lower, electronic (impulse) timers are therefore used. These initiate and terminate the exposure while the ac voltage magnitude is substantially zero. Consequently, an integral number of full *pulses* of current flows through the x-ray tube. Such a timer is hence called an "impulse" timer. The exposure times can only be multiples of 1/120 second, for 60 cycle current: 1/120, 1/60, 1/40, 1/30, 1/20, 1/15, 1/12, 1/10 second, etc.

For even smaller time intervals more elaborate circuits are required, which start and end the exposure at times when the transformer voltage is nearly the peak value. Such special timer units are needed in very rapid sequence work, such as ciné-radiography. For such very short exposures new three-element x-ray tubes are

preferred ("grid-controlled tubes"). Alternatively, the use of three-phase and constant potential voltage generators greatly simplifies the problem of short interval exposures. Such machines are fairly common in Europe, but their greater expense has so far limited their application in the United States.

PHOTOTIMING. We have so far discussed timers which are set by the operator to expose the film for a given over-all time or number of voltage pulses. This means the result is dependent on the accuracy with which the technologist can estimate the length of exposure required. In high kilovoltage radiography times are quite short, so manual timing selection becomes difficult, and an automatic arrangement is needed. "Phototiming" was invented by Morgan in 1942 and is used extensively in almost all radiography, particularly fluoroscopy spot film work and chest radiography.¹⁰

Figure 10 shows the essentials of the method. The exposure setup is conventional except for changes behind the cassette and in the contactor energizing circuit. X-rays transmitted by the cassette strike a fluorescent screen F, whose resulting fluorescence produces a correspondingly large or small current in a photomultiplier tube (P). This current is used to charge a condenser in the phototimer circuit to a predetermined voltage. When this is reached, a thyatron tube activates a relay which terminates the exposure. Thus,

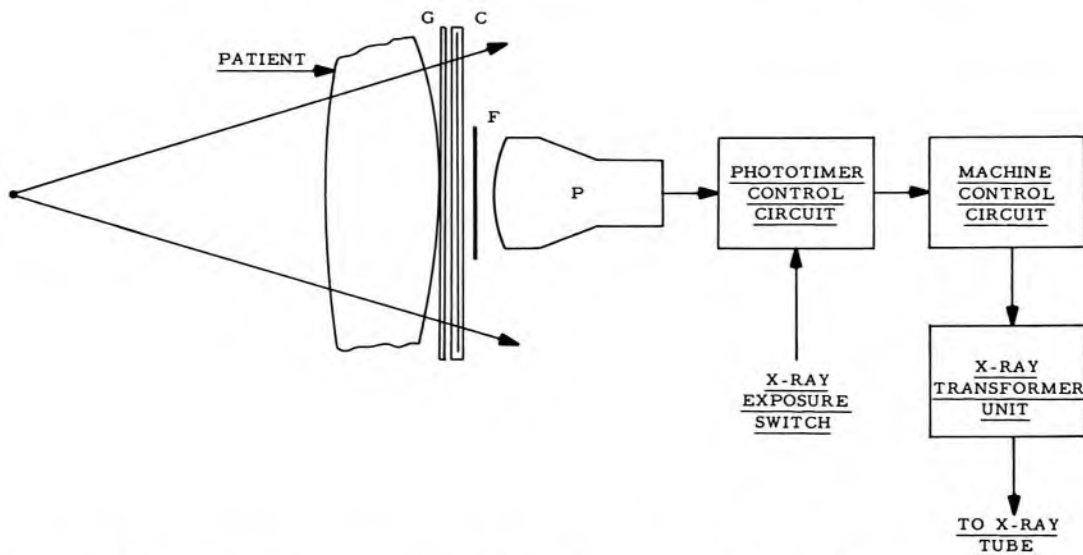


Fig. 10. Phototiming system—principle of operation. X-rays emerge from the cassette (C) rear and cause the fluorescent screen (F) to glow. The photomultiplier tube (P) responds to this glow to produce a current which charges a capacitor in the phototimer control circuit. After accumulation of a certain total charge depending on the desired film dose, the circuit triggers a thyatron which deactivates the contactor switch, turning off the x-rays.

the operator *starts* the x-rays, and the photocell and phototimer setting decide when to turn them *off*, on the basis of how much x-ray exposure penetrated the cassette and hence exposed the screens and film.

Phototiming is extremely useful in applications when essentially the same type of study is repeatedly performed, as in chest and spot filming. Under these circumstances, a fairly constant sample area of the beam may be used provided the patients are positioned reasonably reproducibly. Chest radiography requires that the screen sample the same relative amount of lung and hilar fields, or variable film density will result. This ordinarily presents no serious problem, but extensive lung pathology can affect a particular study.

SEQUENTIAL RADIOGRAPHY. Up to now we have considered primarily conventional procedures, in which rather slowly progressing phenomena are studied. In many so-called "special procedures" one desires to expose radiographs in rapid succession to reveal the rapid dynamics of a physiologic process or movement of opaque material in body cavities or circulation. Older units employ rapid cassette-changers and roll-film units; newer machines, electronic units with ciné or television attachments.

With the development of the image intensifier tube by Coltman in 1948, as many as 60 pictures per second by ciné techniques or television became a technologic possibility. Although resolution is less than that achieved with film changing units, other practical advantages are enormous, and image intensifiers have become an indispensable part of the modern x-ray diagnostic facility. In addition to increasing speed and versatility, the image intensifier operates at considerable less patient dose per picture. A general discussion of the image intensifier and other electronic systems is given below.

ELECTRONIC IMAGING SYSTEMS

Radiologists recognized the limitations of simple fluoroscopy and radiography over twenty years ago. With increasing frequency either brighter fluoroscopic images or very rapid sequence radiography, or both, was required. Fortunately, rapid technologic and scientific progress during and following World War II provided solutions in two devices: the image intensifier tube and modern television systems. These have since been used separately and in combination for diagnostic radiology work.

The radiologist is confronted by a bewildering choice of commercial systems with various features and designs. In this section

TABLE 2. EVOLUTION OF DETECTING SYSTEMS USED IN X-RAY DIAGNOSIS

A. Nonelectronic, Basic Procedures

- | | |
|----------------|--|
| 1. Fluoroscopy | Simple direct viewing after visual accommodation |
| 2. Radiography | (A) direct film exposure
(B) cassettes with intensifying screens
(C) photofluorography |

B. Simple Sequence Radiography

- | | |
|-------------------------|---|
| 1. Direct | (A) rapid cassette changer
(B) roll-film cassette unit (Schonander type) |
| 2. Of fluorescent image | (A) 70 mm Odelca cameras
(B) cinéfluorography |

C. Sequence Radiography with Various Electronic Image Intensifiers*

- | | |
|---|--|
| 1. Full fluorescent image as starting point ¹¹ | (A) Marconi system with special mirror optical system + image orthicon TV.
(B) Cinelix system with special light image intensifier tube |
| 2. X-ray image intensifier tube as starting point ^{12, 13} | (A) use of intensifier tube image for
(1) direct view with mirror
(2) ciné
(3) spot filming by camera unit
(B) Either vidicon or orthicon TV pickup, with usual TV options |

* Note: At this writing, U. S. manufacturers stress x-ray image intensifier systems.

we shall briefly describe the basic components and functions of several systems and then indicate typical combinations and their use. We shall follow a roughly historical approach and avoid insofar as possible specific technical details described elsewhere.

Table 2 briefly summarizes all the basic x-ray detecting systems used in x-ray diagnosis. The first two groups (A and B) include the older nonelectronic types. First are ordinary fluoroscopy and radiography, both previously discussed. The second group involves sequence radiography—that is, obtaining a series of radiographs in rapid succession. The first two types of these move the detectors directly, either the entire cassette or the film itself. The other two involve photography of a large, fixed fluorescent screen, one with a

70 mm camera of great light-gathering power, the other with a movie camera.

All these methods have serious limitations in use. Cassette changers are limited in speed by the inertia and fragility of cassettes, and roll film systems are limited in speed and have other practical disadvantages for many applications. The two final methods of Table 2 (B2) require excessive patient dosage to achieve reasonable image density and quality because so little light from the fluorescent screen reaches the film.

Fluorescent Screen Systems (C1 in Table 2)

The inadequacy of simple methods led to a search for ways to obtain brighter images from those of ordinary fluorescent screens. When modern television became available during the 1940's, Morgan used an image orthicon pickup tube to view the fluorescent screen, displaying the final image on a television monitor. Unfortunately, too little of the light reached the pickup, and the image quality was poor.

More advanced optical designs now, however, incorporate this basic principle in commercial and research instruments (Table 2, C1).¹¹ The *Marconi* system introduces an elegant optical system of modern mirror optics, collecting enough of the fluorescent screen-light for the television pickup to yield acceptable final image quality. The resulting signal can be used in any one of the many versatile television system options, such as:

1. Direct viewing on a monitor.
2. Magnetic tape recording for later rerun.
3. Kinescope film recording from television monitor.
4. "Spot-films" from television monitor.
5. Other arrangements.

The Cinelix system introduces a new electronic element: a *light-sensitive* image intensifier tube (Oude Delft), similar to the *x-ray-sensitive* image intensifier tube to be described below. (Electrons are emitted from a light-sensitive surface and used to produce a new, smaller, much brighter image, by both high voltage acceleration of photoelectrons and image minification.) This tube replaces the television system and yields a bright image which can be directly viewed for fluoroscopy, used for cinéradiography, or televised.

X-ray Intensifier Systems (C2 in Table 2)

Since both of the above systems show the full fluorescent screen area, they can sample areas larger than the 9" or 6" diameter limit of alternative systems. Nevertheless, they are not at this writing very widely employed in the United States and will not be

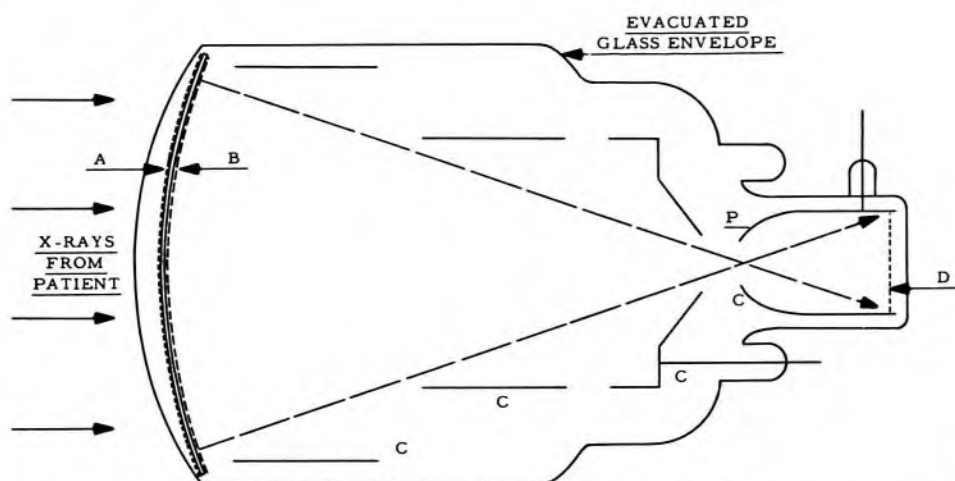


Fig. 11. X-ray image intensifier tube—basic principle (see text). X-rays strike thin transparent plate behind tube face. Surface A is a coating of fluorescent material which glows when irradiated. The resulting light traverses both A and the thin glass plate and strikes the thin, transparent photoemissive surface B on the other side, where electrons are produced in proportion to the light emitted by A. Thus, the x-ray image first produces a *fluorescent image* on A, which in turn produces a corresponding *emitted electron image* on B. A high potential between electrodes B and D drives electrons into a transparent fluorescent screen D at the narrow end of the tube. The electron beam (dashed lines) is focused by electrodes shown (C), as by a simple lens. The electrons from B are accelerated by the high voltage and converge on a small “phosphor” D. Intensification results from two facts: the image is reduced in size (minification brightness increase); and the electrons producing the image at D have very high energy because of the high accelerating voltage (about 25 kV).

discussed further. Most installations employ the *x-ray image intensifier tube* as a basic constituent (Table, C2). Invented by Coltman in 1948, it obtains a bright image *directly* from the x-ray beam using a single tube.^{12, 13} It is shown schematically in Figure 11. The image intensifier consists of an evacuated glass bottle with several components, indicated in part by A, B, C, and D.

DESCRIPTION. A is a zinc-cadmium sulfide fluorescent screen, especially designed to yield maximum brightness with reasonable image sharpness. In rather close optical contact with the fluorescent screen is a transparent photoelectric surface B (in some models a very thin glass sheet separates A and B, supporting them both), of multialkali design, emitting electrons from local areas in proportion to the light intensity striking them. Hence, photoelectrons are produced by B in a pattern corresponding to the x-ray intensity pattern on A. B is maintained at a high negative voltage relative to anode D, so the photoelectrons are accelerated from B toward D. They traverse the hole in P and strike the electrically conducting fluorescent

screen. The electrodes C are designed to focus the electron beam like a lens, so D glows in a pattern corresponding to the photoelectric current from B, and hence the original x-ray intensity pattern.

The high voltage accelerates the electrons to substantial speeds, so the "output phosphor" D glows quite brightly. This alone contributes a factor of the order of 15 times increase in image brightness. In addition, the electrons' energy is concentrated over a reduced area, so a brightness increase results from *minification* of the image. For example, a 5" diameter tube yielding a 1" diameter output image has an intensification of $5 \times 5 = 25$ times from this cause alone, yielding $15 \times 25 = 375$ as a total factor. On the other hand, a 9" tube correspondingly yields an 81 times factor, with $15 \times 81 = 1,215$ total. In general, therefore, the intensification factor depends not only on the tube design but also on its operating voltage and dimensions. Since electron focusing is efficiently carried out for only a limited range of operating voltage, the tube intensification factor is not normally very adjustable.

APPLICATIONS. The simplest image intensifier system uses no other electronic devices; its application is limited to direct viewing of the output phosphor by mirror system (fluoroscopy) or ciné. According to Morgan, the former may leave much to be desired because so few light photons are transmitted to the eye by most mirror systems.¹⁴ Consequently, the visual image is ordinarily not improved much over direct fluoroscopy. Television viewing systems are for this reason often preferred for fluoroscopy with image intensifier tubes. However, it is quite useful to obtain ciné or spot films (either single ciné frames or 70 mm film) of the output phosphor image. Such uses are well-established as yielding worthwhile results.

Fluoroscopy is thus preferably carried out by adding a television pickup and visual monitor unit to improve the visual image. This of course also makes possible the use of other television gadgets such as tape recording, kinescope recording (photography of the monitor screen, usually with 70 mm film), etc. These all add to the equipment cost, and the television monitor screen image is inherently poorer in resolution than the intensifier tube output phosphor image.

The design problems of electronic image intensification are many, and the selection of a unit for purchase is a complex choice involving the specific application anticipated as well as many rapidly changing technological designs. Studies are in progress to evaluate the most useful applications of these devices and the relative performance of alternative designs. Hopefully, specific design and application recommendations will become available in the near future. In the meantime, the basic usefulness of electronic imaging systems

for an ever increasing number of diagnostic procedures is not questioned.

Can electronic imaging systems reduce the patient dose in x-ray diagnosis? The answer is probably affirmative but to a relatively limited extent. Acceptable information yield requires x-ray photons in adequate numbers. Too fast a system always has much quantum and other noise, with excessive image degradation. The contribution of these systems, therefore, is primarily in making new studies possible without prohibitive patient and operator dosage. These studies yield diagnostic information never before obtainable, opening new horizons for x-ray diagnosis.

X-RAY FILM

Perhaps the most crucial tool in radiography is x-ray film. This is the repository of the final image, lasting and available for future re-examination and comparison. Film is very versatile, with a great many available types and applications. We shall consider the basic radiographic types, pitfalls in their use, and basic film characteristics.

Film Types (Table 3)

X-ray and photographic films generally consist of a plastic base coated on at least one side with an emulsion containing silver bro-

TABLE 3. SOME BASIC FILM TYPES USED IN X-RAY DIAGNOSIS¹⁵

General Type	Application
1. Direct action non-screen	Fine detail work: breast, extremities in medical work; industrial radiography.
2. Screen type films, used with intensifying screens	General medical use: faster, with moderately good detail. Used to minimize motion blur.
3. Light-responsive, single-emulsion film	(A) 70 mm photofluorography and kinescope spot films. (B) spot films, of image intensifier tube. (C) ciné, of image intensifier tube.
4. Polaroid process	Rapid processing system, useful in operating room orthopedic work.

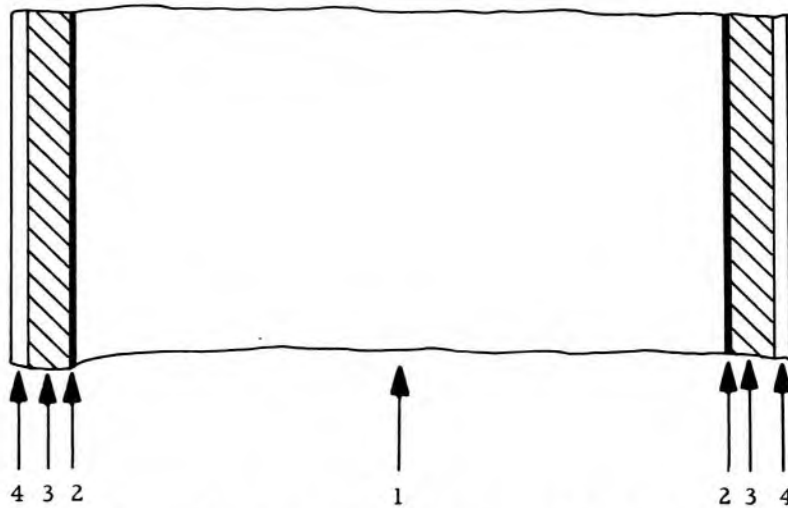


Fig. 12. Cross section of a two-emulsion x-ray film. 1. Film base of cellulose acetate or polyester, about 8 mils thick (1 mil is 0.001 inch). 2. Adhesive. 3. Emulsion: silver bromide plus gelatin plus other special ingredients. Thickness $\frac{1}{4}$ to $\frac{3}{4}$ mil, depending on design. 4. "T-coat:" about 0.2 mil thick surface protective coating. (From data courtesy of Eastman Kodak Co.¹⁵)

vide.¹⁵ X-ray films are usually coated on both sides for increased sensitivity (Fig. 12). Screen type film has a relatively thin emulsion to assure good resolution; it is activated primarily by intensifying screen fluorescent light which does not penetrate a thick emulsion very well. Single emulsion films are used essentially photographically to record various phosphor images. They are less sensitive than double-emulsion films but have potentially greater resolution.

Direct action film emulsions may be increased in thickness and silver content for enhanced sensitivity to x-rays; this is done in medical no-screen films. Resolution is reduced in the process, however, and finer detail direct action industrial films have much thinner emulsions.

Polaroid films have the advantage of rapid development and have found wide use in general photography. A major application in medical radiography is to orthopedic work, employing a special exposure-development unit with its own built-in intensifying screen. Its rapid processing is very useful in some work, but limited image quality at present restricts the application of Polaroid radiography.

We shall confine our further discussion to the ordinary two-emulsion films used in medical and industrial radiography. Two aspects will be considered: practical ones relating to pitfalls in use, particularly in more demanding applications, and basic film characteristics.

TABLE 4. PITFALLS IN USE OF X-RAY FILMS

Storage

1. Even when sealed: radiation fog, heat deterioration.
2. When unsealed: chemical vapors, relative humidity also.

Handling Artifacts

1. Pressure: *before* exposure, *light* mark on film.
after exposure, *dark* mark on film.
2. Static electricity: tree, crown, and smudge types.
3. Light leaks (storage): mottled darkening of exposed areas.

Film Processing

1. Development: (nonreproducible density, fog, streaking, etc.).
(A) Solutions: strength fluctuation with use.
impurities: mercury, copper, sulphur, fixer, etc.
(B) Procedures: time and temperature; agitation; quick film transfer;
preliminary stirring, etc.
2. Rinsing: at least 30 seconds; strong acetic acid stop bath for exacting work.
3. Fixing: frequent solution changes; agitate first minute; leave in twice clearing time.
4. Washing: not too cold, *turbulent* water, at least 30 minutes.
5. Drying: detergent dip first; then use dry heat, with air circulator.

Pitfalls in Use (Table 4)

As in the use of any tool, x-ray films must be employed with reasonable care to assure good results. Otherwise, problems can arise in obtaining reproducible film density and contrast. Fog and artifacts can obscure images if care is not taken during the storage, handling, and processing of films. Problems of quantitative film dosimetry are particularly great and are considered separately in Appendix B.

STORAGE AND HANDLING. Careless storage of films can result in their premature deterioration. Even when kept sealed in the box they can be fogged by ionizing radiation. Films with high silver bromide content, such as medical no-screen x-ray films, are especially susceptible. (These normally deteriorate with less than twelve months storage.) Light-sensitive films are far less sensitive to stray ionizing radiation. Normally medical screen films may be stored in a sealed

box at moderate temperature for as long as six months beyond their stamped date. Once the box is unsealed, however, chemical vapors like ammonia, hydrogen sulfide, formalin, etc. take their toll. Prolonged storage should be avoided. Deterioration is particularly rapid at high temperature and relative humidity.

Artifacts can arise both before and during processing. Film processing artifacts are due to deteriorated or contaminated solutions and are normally not a serious problem with competent technologists. Nonchemical artifacts occur occasionally, however, in even the best-run department. They are of three types: pressure, static, and light artifacts.

Pressure artifacts arise when the film is crimped or pressed by a hard, sharp object during handling before development. A telltale mark results. It is white if the stress occurs before, black if after film exposure. Static electric discharges activate the film locally along and near the discharge path; too rapid film handling in dry weather is usually responsible. Humidity control is sometimes desirable in the film-loading room to minimize this problem. Finally, since x-ray film (particularly screen film), is light-sensitive, light leakage of a film storage bin is to be avoided.

FILM PROCESSING.¹⁶ Film processing is a complex conversion of the invisible "latent" image of the exposed film into a visible, permanent record. It must be carried out properly to avoid image deterioration. A brief review of the steps in film processing is included below to indicate the reasonable but crucial steps involved.

The original exposure of film to light or x-rays results in reduction of silver salt to silver at discrete crystal locations. (This reduction is quite irreversible when caused by ionizing radiation; when caused by light the effect is modified by many factors. Consequently, the latent images differ basically. These differences give rise to differences in density and "reciprocity-law" characteristics discussed below.) Remarkably, very few such "active centers" are produced by either x-ray or light, and negligible silver would be deposited if the unaffected emulsion were removed without development.

The developer is an alkaline solution which very effectively reduces either part or all of each crystal containing a reasonable number of activated silver atoms. In the process, for each reduced silver atom originally present, a "fast" developer reduces all the silver ions in the associated crystal, yielding a single *silver grain*; up to a *million-million* silver atoms may result per atom originally activated! The sensitivity of film is thus enormously increased. On occasion it is desired to keep grain size small for better resolution. "Fine grain"

developers, which develop only a *part* of the AgBr crystal, are used, so smaller grains result; the total silver, however, is correspondingly reduced, and the final film is much lighter. During development some reducing action of *unactivated* crystals inevitably occurs. This produces "development fog," which increases with development time and temperature.

Since development is a chemical affair, it depends strongly on developer solution concentration and pH as well as on impurities which affect chemical action. Careful chemical replenishment is needed to maintain solution concentrations reasonably reproducible. Oxidation occurs at the tank top, and preliminary stirring is needed to prevent so-called "air-fog." Solution temperature is crucial. Development time can fortunately be varied to compensate for temperature changes. Between 64-70° F., such compensation assures reproducible film density with negligible loss of film contrast. Finally, chemical impurities can be disastrous in effect. These include sulfur, mercury, and copper as well as fixer, acids, etc. Special procedures are needed in hand developing high detail and high contrast diagnostic and dosimetry films; they are discussed separately below.

The developed image is rinsed before fixing. Rinsing stops development at the desired time and removes chemicals from the film to prevent fixer deterioration. Ordinarily, a 30 to 60 second immersion in water is adequate. For more exacting applications a high acidity acetic acid stop bath is used (pH — 2.5 to 4 *maximum*). The films are transferred quickly from the developer without draining, to minimize contamination and facilitate replenishment of the developer solution.

The fixer makes the image permanent in two ways. It removes the undeveloped silver bromide, thereby preventing later film darkening from the action of light, and it hardens the emulsion. Fixer solutions should be replenished quite frequently to maintain their strength.

After fixing, films must be adequately washed to avoid deposits of chemicals. Finally, drying should be properly carried out, as indicated in Table 4.

What of the role of automatic film processors? Their great contribution is their speed and minimization of the human factor, assuring fair to good quality development of routine roentgenograms. They are invaluable for the vast majority of studies, especially in busy departments. In high contrast, fine detail studies such as mammography, however, two problems arise. First, such machines are not usually designed to transport thicker emulsion films without substantial modification. In addition, no automation has so far com-

pletely solved problems of developer speed variation and solution deterioration, important considerations in studies when reproduction of film density is very important.

Basic Film Characteristics

We have thus far spoken about the structure and use of films but not their exposure characteristics. We shall now briefly discuss the two basic types of film response to radiation, as well as some other properties of practical importance in radiography.¹⁷

FILM DENSITY. When films are exposed and developed, the various areas are darkened in response to the x-ray intensity striking them. The degree of film darkening is generally referred to as the "film density." More precisely, film density (D) is defined in terms of the transmission of viewbox light (T):

$$T = 10^{-D} \quad (8-3)$$

For example, a film density of 1.0 yields light transmission values through the film of $10^{-1} = 0.1$; of density 2, .01; of density 3, .001. In ordinary clinical work, useful film densities range from 0.3 up to 1.7 or so, permitting the use of an ordinary viewbox for most work. Densities of much greater than 2 require brighter viewboxes (100 times brighter for $D = 3$ than for $D = 1$!), or stray light and glare excessively reduce the retinal contrast.

TABLE 5. COMPARISON OF REQUIRED EXPOSURES OF SCREEN-FILM AND DIRECT FILM EXPOSURES TO PRODUCE VARIOUS DENSITIES¹⁵

Film Density-D	Light Transmission-T	Corresponding Relative Film Exposures (that for $D = 1$ taken as unity)	
		Typical Screen-type Film with Par Speed Screen	Direct Exposure Film
0.4	0.4	0.45	0.4
0.7	0.2	0.75	0.7
1.0	0.1	1.0	1.0
1.3	0.05	1.3	1.3
1.6	0.025	1.65	1.6
2.0	0.010	2.2	2.0
3.0	0.001	6.0	3.0

Note: The direct exposure film density changes proportionally with exposure (first and fourth columns). However, screen film departs somewhat from proportionality at densities of 0.7 and 1.6 and is very slow above a density of 2.0 (first and third columns).

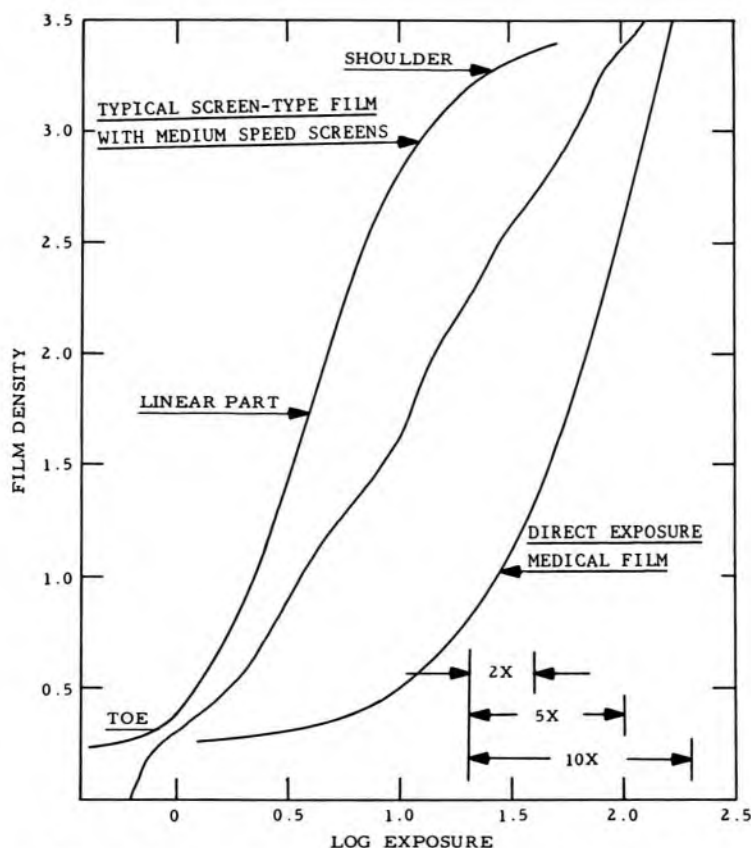


Fig. 13. Basic film characteristic curves (H-D curves).¹⁵ At left, typical screen-type film, exposed with medium speed intensifying screens. At right, direct exposure medical film. Note the marked difference in curve shape. This is because the screen film is activated primarily by light, the direct exposure film by x-rays. Separate horizontal scales are used because screen-film speed varies greatly with factors which do not affect directly exposed film response. The numbers at lower right indicate the factor of increase of exposure for indicated distances along log exposure scale. (From data courtesy of Eastman Kodak Co.¹⁵)

CHARACTERISTIC CURVES. The film "characteristic curve" is a graph of its film density with standard development versus the *logarithm* of the exposure. First described by Hurter and Driffield in 1890, this type of curve is usually called an "H-D curve." Figure 13 shows representative H-D curves for two film types. (Table 5 compares the film density, light transmission, and required relative film exposures for a par speed screen-film combination and medical no-screen film. That for 1.0 density is taken as 1.0 in both cases.) There are three parts to the H-D curve: the toe, straight part, and shoulder (Fig. 13). With *all* films, the curve reaches a peak at very high exposure levels (not shown), then reverses more or less symmetrically (the basis for "solarization" techniques).

One might reasonably ask why is D plotted versus $\log E$ instead of E ? There appear to be two reasons. First, a logarithmic scale permits including a wide range of exposure on the same graph, which

is useful in showing the complete response curve of even a single film as well as in comparing films of different speeds. In addition, the slope of the H-D graph directly yields the film contrast in diagnosis (see below).

COMPARISON OF DIRECT- AND SCREEN-EXPOSED FILM RESPONSE. From Figure 13 and Table 5 it is evident that screen exposure H-D curves differ greatly from direct x-ray exposure curves. It is interesting that *all* films exposed to light have curves generally shaped like that of typical screen-type films exposed with intensifying screens. Of course, the slope and sensitivities, as well as grain size, will differ. Also, all directly exposed films—*even those designed for intensifying screens*—have curves like that of non-screen film! Evidently, the basic curve shape is characteristic of the energizing radiation—i.e., whether light or ionizing radiation. It is also interesting that films exposed directly to x-rays follow the reciprocity law (see below) whereas those exposed by light do not.

Direct exposure films include Kodak medical No-Screen and AA- and M-Industrial types. Screen-type films include Blue Brand, Royal Blue, and F-Industrial. (Equivalents of other manufacturers exist.) Direct exposure films are used in industrial radiography at high energy with thin lead screens, which serve as intensifiers by adding photoelectron irradiation to the emulsions.

Returning to Figure 13, screen films exhibit a rather short toe region with a straight, steep linear part from about 0.6 to 2.2 density, beyond which the curve curves and flattens out. The directly exposed films have a quite long toe region, beyond which curve steepness increases steadily. Curves of films designed for direct x-ray exposure do not flatten out until densities well in excess of 3 are reached because the emulsion silver content is great. As a result of these facts, screen films are ordinarily exposed to obtain final densities of about 1.0 in regions of major interest and viewed with ordinary viewboxes; direct exposure films are best exposed to densities from above 1.5 to well over 3.5 and viewed with special industrial high intensity viewboxes.

Table 6 compares the approximate speed and film contrast of screen-type film using par speed screens with those of direct exposure and faster and slower industrial films. The screen film when used with screens is of course much faster than the others. In addition, note that its contrast increases only very slowly with film density. Contrast of the directly exposed films, on the other hand, increases roughly in proportion to the film density. Screen exposures yield good contrast at low film densities at which the other films have much lower contrast, but the direct exposure films rapidly catch up and exceed screen film contrast at higher densities.

TABLE 6. COMPARISON OF CONTRAST AND RELATIVE SPEED OF FOUR FILM TYPES EXPOSED TO THREE DENSITY VALUES¹⁵

Item	Typical Screen-type with Par Speed Screens	Direct Exposure	Faster Industrial	Slower Industrial
Relative Exposure to Produce Density of:				
D = 1.0	0.08*	1.0	6.85	26.3
D = 1.5	0.07	1.0	6.40	24.2
D = 2.0	0.07	1.0	5.95	21.6
Film Contrast for:				
D = 1.0	2.65	1.70	1.92	2.13
D = 1.5	2.93	2.77	3.22	3.46
D = 2.0	3.33	3.18	3.98	4.42

* Very approximate data; dependent on type and particular screens used, exposure times, and screen temperature.

RECIPROCITY LAW.¹⁸ The reciprocity law states that the response of a film to a given total exposure of a certain kind of radiation is independent of how fast the exposure was administered. For example, if the law is valid a 100 mA exposure of 1 second should yield the same film density as a 500 mA exposure of 1/5 second, with other factors identical.

The reciprocity law holds quite well for films exposed directly to ionizing radiation. It does *not* hold, however, for those exposed primarily to light, as in screen techniques. For such techniques, there is usually a certain exposure time for which a minimum total mAs exposure is required, with greater values needed for either longer or shorter times. About twice as much mAs is required for a 10 second as for a 0.1 second exposure.

This change in sensitivity with exposure time is not very troublesome in clinical radiography because exposures are normally well within the range of 10 to 0.01 second. In industrial radiography, however, intensities change greatly when distances are varied in the course of radiographing different objects, and fairly long exposure times are often used. Simple inverse square law corrections are usefully made for direct film exposures but may be quite inaccurate for some fluorescent screen work.

DIRECT EXPOSURE FILMS—CONTRAST, SPEED, AND RESOLUTION. We now add a few comments about film contrast, speed, and resolution with direct exposure technique. The questions have been raised in mammography: "Can the graininess of faster films be improved without reducing their speed, or the speed of slower films improved without increasing their graininess? And, what are the contrast potentialities of fine grain films?" These questions arise in studies in which a high visual stimulus with maximum resolution is required to facilitate detection of tiny objects. It will be recalled this demands high contrast with minimum motion blur and mottle noise, so that fast, fine grain film is ideally preferable.

The characteristic curves of directly exposed films tend to parallel each other, especially with full development. They therefore have generally similar basic contrast properties, and production of higher contrast films appears unlikely at present. Fortunately, even existing direct exposure films provide considerable contrast when exposed to higher density. Since directly exposed films have linear density versus exposure curves, it can be shown that the luminance contrast of the film as seen on a viewbox is enhanced $2.306D$ times, where D is the film density. Thus, factors of nearly 7 times are achieved in contrast enhancement at density values of 3 (Fig. 14). This principle has been routinely applied with profit for many years in industrial radiography.

What of film speed and resolution? Film speed can be increased in at least three basic ways: the use of more silver bromide in the emulsion, larger crystals, and fuller crystal development. Unfortunately, all these increase film graininess, and increasing speed worsens quantum mottle; hence, increased speed works both ways to *reduce* resolution.

It therefore appears unlikely at present that faster films of high quality resolution can be developed because of the inherent nature of the photographic process. Contrast enhancement is best achieved in the exposure itself, and no film improvements are currently envisaged which will substantially increase film contrast.

SPECIAL DIAGNOSTIC TECHNIQUES

Considerable ingenuity is required to obtain maximum useful information from an x-ray examination of a patient. One desires roentgenographs of *specific objects* with good image quality. For a single object in a uniform body this involves only obvious contrast and resolution considerations. In human roentgenography, however, many different objects are simultaneously traversed by the same

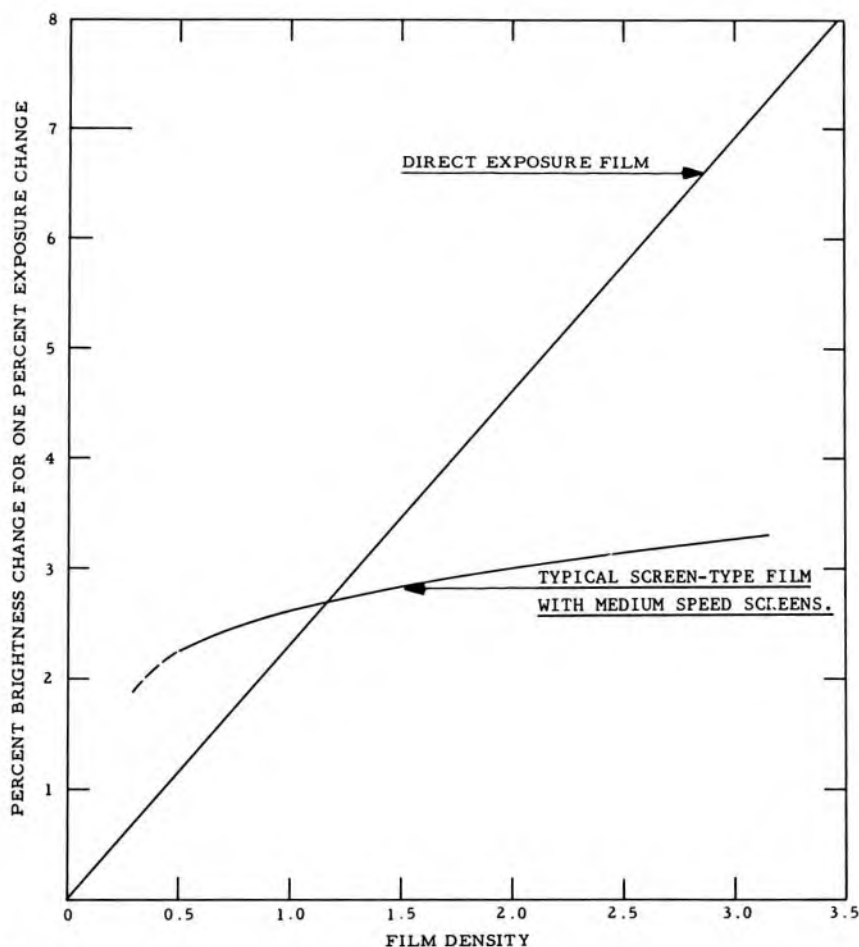


Fig. 14. Luminance contrast vs. film density, for screen vs. direct exposure films. Films viewed on illuminator. Note contrast of intensifying screen film does not change much with film density whereas that with direct exposure technique rises in proportion to density, attaining rather high values.

x-ray beam, and often parts of diagnostic interest are obscured by others. To illustrate, in Figure 15 the heart obscures the image of the nodular lesions N in the PA view (solid lines). A similar nodule at M would be more readily observed.

Two basic approaches exist to highlight these desired objects. The first is to selectively increase the contrast of the object of interest in some way; this, of course, is the reason for special contrast studies. The second is to employ selective views to minimize the obscuring action of other anatomic objects. This section briefly indicates useful techniques employing both approaches with a few additional comments regarding enlargement techniques.

CONTRAST ENHANCEMENT. Basically, one can increase contrast in the x-ray beam itself, in the detector, or in a new image derived by a "dodging" technique.

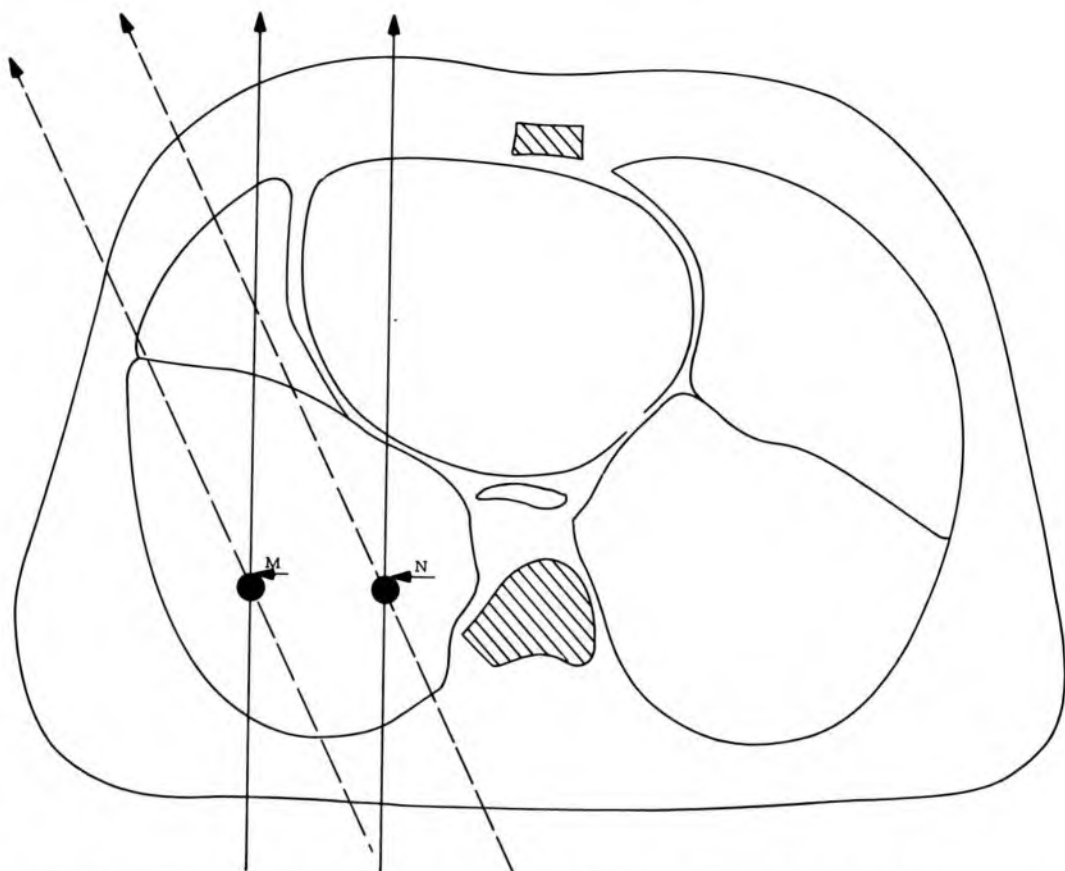


Fig. 15. Usefulness of multiple views. Other objects behind or in front of a lesion can lower its detectability by both superimposing confusing detail and reducing contrast. For example, consider identical lung nodules N and M. Both contribute the same inherent x-ray contrast, having the same $(t\Delta\mu)$ values. However, in the ordinary posterior-anterior (PA) projection (solid arrows), the image of N is obscured by the relatively dense heart, that of M only by aerated lung. As a result, the film density of N's image is much less than that of M, with greatly reduced final contrast. In the oblique view (dashed arrows) images of M and N are more nearly the same in quality.

The beam contrast originates in the difference in attenuation of the object and its surrounding material; this is always reduced by scatter fog. Low kilovoltage techniques are helpful in highlighting components of atypical composition like bone, fat, calcific deposits, etc. Low beam penetration, however, then limits transmission. Hence, low kilovoltage techniques are limited to thinner parts like extremities. A lower kilovoltage limit is set by required exposure mAs values; if these are excessive, both motion blur and tube anode overheating can become problems. Dosage to superficial parts of the patient also can become unacceptably great with beams of very low penetration.

Contrast media give considerable information obtainable in no other way. Generally objects are visualized which would otherwise

not be visible even without obscuring objects present. In addition, if the resulting opacity is high, any other objects in the beam will not obscure the image to the same degree because the desired subject contrast is so high. This is the reason high iodine content contrast media are so useful in visualizing blood vessels.

Scatter reduces overall roentgenogram information and is normally controlled by the use of grids and minimum size fields as previously described. In some studies an "air-gap" technique is also used.¹⁹ Figure 16 shows chest techniques, both ordinary (film at A) and air-gap (film at B). Fewer scattered photons reach the film in position B and A because many photons miss the cassette altogether at the remote location. Air-gap technique is used for high kV chest work with a long TFD (6 to 10 feet), to reduce magnification and penumbra problems. A gap of about 6" is often employed. Air-gap is also an incidental benefit in enlargement or modification technique.

We have referred previously (p. 263) to the contrast enhancing action of film and television systems.

Dodging is a technique of deriving a new image with more suitable contrast characteristics from roentgenograms. It is used either

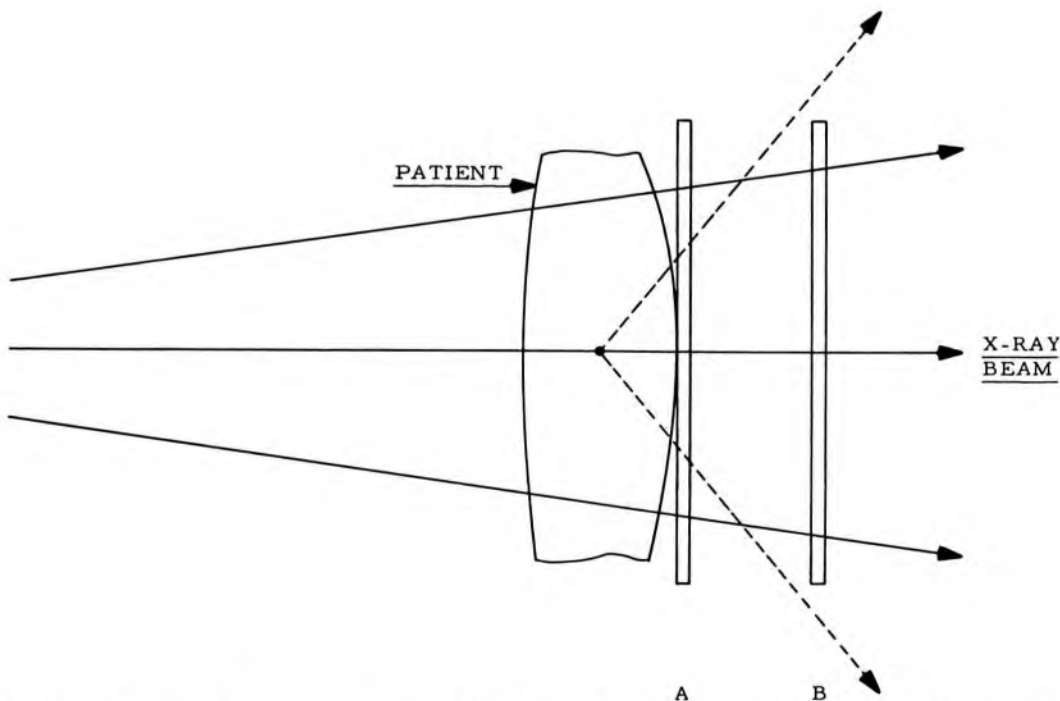


Fig. 16. Ordinary vs. air-gap technique for chest radiography. A. Ordinary: cassette relatively close to patient, ordinary TFD values. B. Air-gap: cassette 6 inches further from patient. Magnification and penumbra and controlled by longer TFD (6 to 10 feet). Note that some scattered radiation hitting the cassette in A position (dotted lines) misses it entirely in B.

TABLE 7. SOME CONTRAST "DODGING" METHODS

Definition	
Using one or two ordinary roentgenograms to obtain a new image with more useful film density distribution. Purpose: to increase either contrast or latitude.	
Methods	
(A) Photographic: special film copying techniques.	
(B) Electronically: use of non-linear pickup, amplifier, or monitor system.	
1. Logetronic system	
2. TV system: direct from film on viewbox	
3. TV system: from image intensifier	
(C) Subtraction techniques	
Procedure: combine two roentgenograms of immobilized patient, taken pre- and post-injection of contrast, to highlight vessels and de-emphasize anatomy common to both exposures.	
Three techniques	
1. Photographic: combine contrast roentgenogram on viewbox with <i>negative</i> print of pre-injection roentgenogram, thereby partly cancelling out common anatomy.	
2. Two TV cameras: reverse one TV image electronically and combine images on a single monitor screen.	
3. Color viewing of two roentgenograms: mount films on two boxes, and view one with a red, the other a green, filter. Use optical system to combine the two color images. Since complementary colors are involved, common images show white; the vessels, red or green. Convenient and effective, and relatively inexpensive.	

Inherent Limitation

These methods affect only contrast and cannot add information not originally present. Not substitute for additional views, tomography, etc.

to make a desired part stand out more (heighten contrast) or even out the film density (increased latitude). Photographic and electronic techniques are used for ordinary dodging; "subtraction" is a related technique useful in highlighting contrast media in cardiovascular opacification procedures.²⁰ Although a detailed discussion is beyond the scope of this presentation, Table 7 summarizes the various methods.

It should be stressed that dodging cannot put into the roentgenogram information not already there; it only enhances. Actually, as in all such meddling, some information is bound to be lost. Only additional views can add to the examination information.

Selective Views

As indicated in Figure 15 there is generally an optimum radiographic projection to best visualize a particular object in the body. Of course, economics of expense, time, and patient dosage limit the practical routine number of views, and medical judgment must determine their selection. Normally, posterior-anterior and lateral projections are employed in chest studies, with oblique views often added. However, two types of special exposure techniques are of great benefit in many situations: stereoscopic and body section radiography.

STEREOSCOPIC RADIOGRAPHY.²¹ Implicit in two views with perpendicular projections is the three-dimensional arrangement of the patient's body parts. In practice, however, it is often very useful to obtain a direct three-dimensional illusion to better evaluate the

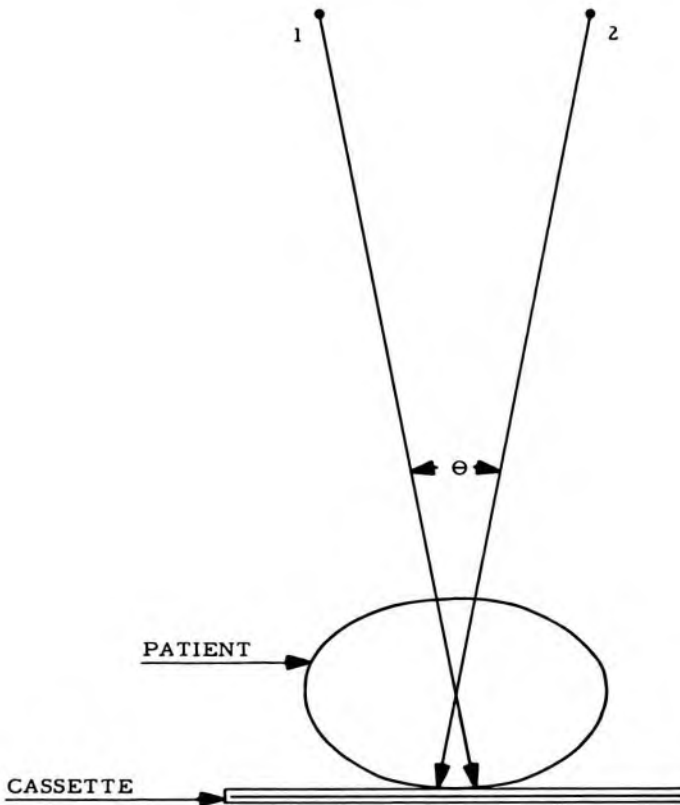


Fig. 17. Exposure of stereoscopic views. Two roentgenograms are exposed. The first is from angle 1, to the left of the central ray; the second, from angle 2, to the right of the central ray. The angular shift θ corresponds to the average interpupillary viewing angle. Later the films are studied simultaneously in a special viewer which facilitates superimposition of the two views, with image "fusion" in the radiologist's brain. A three-dimensional picture results, clarifying the location of various structures otherwise superimposed.

depth of an object or lesion. For such work, two stereoscopic exposures are often made. (Fig. 17). The cassette and patient are fixed in position, but the x-ray tube is shifted through the appropriate interpupillary viewing angle. The resulting two roentgenograms are then viewed by the radiologist using a suitable stereoscopic device, so each film is seen by one eye only, at an angle of view corresponding to the exposure angle shift. When viewed properly, the film images may be "fused" in the radiologist's brain, yielding a three-dimensional image of the patient's anatomy.

For some work stereoscopic views are taken to ascertain the location of a particular object—for example, a foreign body—relative to an anatomic landmark. Figure 18 indicates how one can compute the location of an unknown object X in the body, relative to a fixed anatomic landmark R. (The two formulas are easily applied in practice since factor $(1-k/D)$ and the shift angle are relatively constant in a

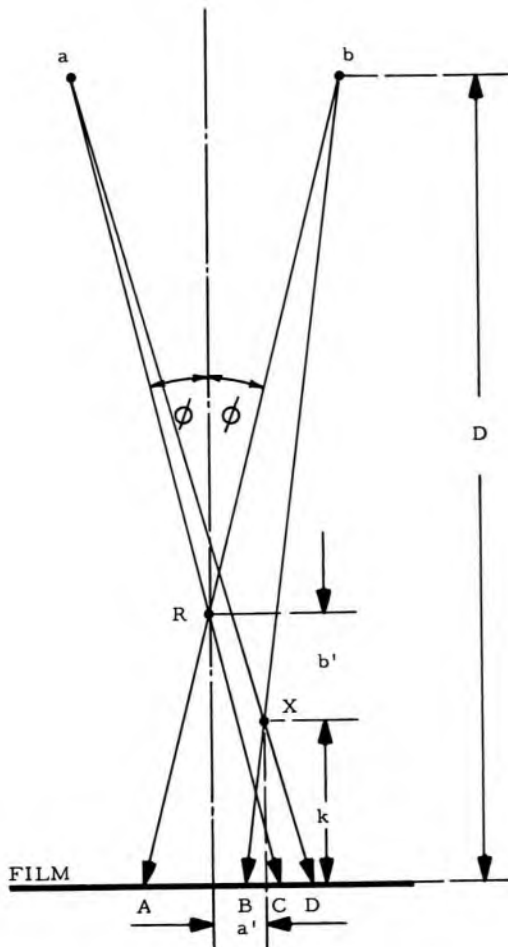


Fig. 18. Foreign body localization formulas. Object: to evaluate how much closer to the film and how far lateral a foreign body X is with respect to an anatomic landmark R. Procedure: first, a direct view is taken to decide on the general location of X. Then, two views are exposed, a and b, displaced in either direction from the direct view, in a plane including points a and b. Measurements are then made on the films of the separations d_1 (AB) and d_2 (CD) of the images of R and X in the two views. The formulas for a' and b' :

$$a' = \frac{d_1 + d_2}{2} (1-k/D);$$

$$b' = \frac{d_1 - d_2}{\tan \phi} (1-k/D).$$

Here, D is the target-film distance, and k is the distance from object to film. If k is large compared to b' , and D is large, any error in estimating k is easily made clinically negligible.

given setup.) Such calculations are useful to yield an accurate position to guide the surgeon and are also helpful for those who do not have stereoscopic vision.

BODY SECTION STUDIES. Body section radiography is used to obtain the clearest possible roentgenogram of structures in a thin plane slice of a patient. The preferred term is tomography, but laminography, stratigraphy, planigraphy, and other names have also been used.

Two basic benefits can be achieved. First, one can minimize the confusion introduced by superimposed images from structures on either side of that under study. These introduce details which at least distract from the relevant images and may actually obscure them. This is particularly true in some skull and chest studies. Second, one can combine a series of tomograms of layers, say, 1 cm apart, to direct specific attention to an anatomic location at a particular depth in the patient. Tomography is used to better visualize brain ventricles and bony structures in the skull, lung lesions, biliary ducts, urologic lesions, etc.

The basic principle employed is that motion blur results not only in broadening of the margins of an image [Fig. 4(Top)], but also in *contrast* reduction. Figure 21 in Chapter 3 illustrates the loss of contrast of several small test objects due to their uniform motion during the x-ray exposure. In general, for uniform motion, it can be shown that the contrast is reduced approximately by the factor:

$$R = 1 + \frac{\text{motion amplitude during exposure}}{\text{object size}} \quad (8-4)$$

For example, in Figure 21 of Chapter 3, the image of a 0.5 mm object is reduced in contrast by factors of 2, 3, and 5 by motions of 0.5, 1.0, and 2.0 mm, respectively. Similarly, the contrast of obscuring images can be reduced in tomography by selectively moving the parts involved.

Figure 19(Top) illustrates how this may be accomplished. Suppose one wishes to demonstrate object A and blur out the image of object B which would normally obscure it. The tube and film are moved in *opposite* directions during the x-ray exposure. This is done in such a way that the image of A is never blurred—i.e., the x-ray image of A travels from Q to P at exactly the film speed. It is easily shown by simple geometry that the image of any other object in the plane through A parallel to the film (“fulcrum plane”) is also sharp.

What of objects away from the fulcrum plane? Consider B, closer to the film than A. Its image moves from Q' to P'—much less than the film motion Q to P. B's *image* is therefore blurred since it is spread out on the film through a distance (PP' to Q'Q). B's *contrast*

is also reduced (8-4). [The amount is determined by its size and the distance (PP' and $Q'Q$), which in turn depends on B 's distance from the fulcrum plane and the total angle of motion during the exposure.]

Thus, tomography substantially reduces the contrast of all obscuring objects on either side while retaining the contrast of those in the desired plane; the contrast reduction *increases* as the image size *decreases*. Figure 19(Bottom) shows how a small image is lost in superimposed detail in an ordinary study. Note how a tomogram in Figure 19 reduces the contrast of the irrelevant detail (a factor of 8 times is shown) while that of the desired image is unchanged. If the object is larger its image contrast is not reduced as much, but the over-all image density made more uniform. The result is a relatively "bland" visual background with the desired image highlighted.

Three basic types of tomography motion are used. The most common uses *linear* motion, such as shown in Figure 19. It is usually easily added to ordinary radiographic machines at reasonable cost. One ingenious version employs "book cassettes." These are multiple intensifying screen units in which screen films are simultaneously exposed parallel to each other (usually 1 cm apart) and to the fulcrum plane. The screens and films and their sequence are arranged so that the remote films are most sensitive and the proximal ones least sensitive, to compensate for x-ray absorption by the screens. The result is a sequence of films exposed to about equal final densities. Since the films vary in distance from the fulcrum, they precisely, simultaneously, and automatically sample different depths in the patient. Book cassettes are very useful in studies in which patient motion between successive exposures could otherwise introduce uncertainty in the selected anatomic locations.

The other two tomography systems are less frequently employed because they require more elaborate apparatus. The first is the "polytome," which has three nonlinear movements: elliptical, circular, and hypocycloid. This is a high precision device for selectively radiographing a very small area of the body, such as the small bones of the inner ear. This device corrects for some deficiencies of the simple linear tomogram, not considered in the above brief discussion.

The second is horizontal or axial transverse tomography. Recall that in linear tomography the image is parallel to the patient's axis because of the way the motions must be carried out. In some situations one requires a section *perpendicular* to the patient axis; this cannot be obtained with linear tomography. Horizontal tomography involves simultaneous *rotation* of the patient and the cassette about parallel axes.

The details of these systems are beyond the scope of this discussion.²²

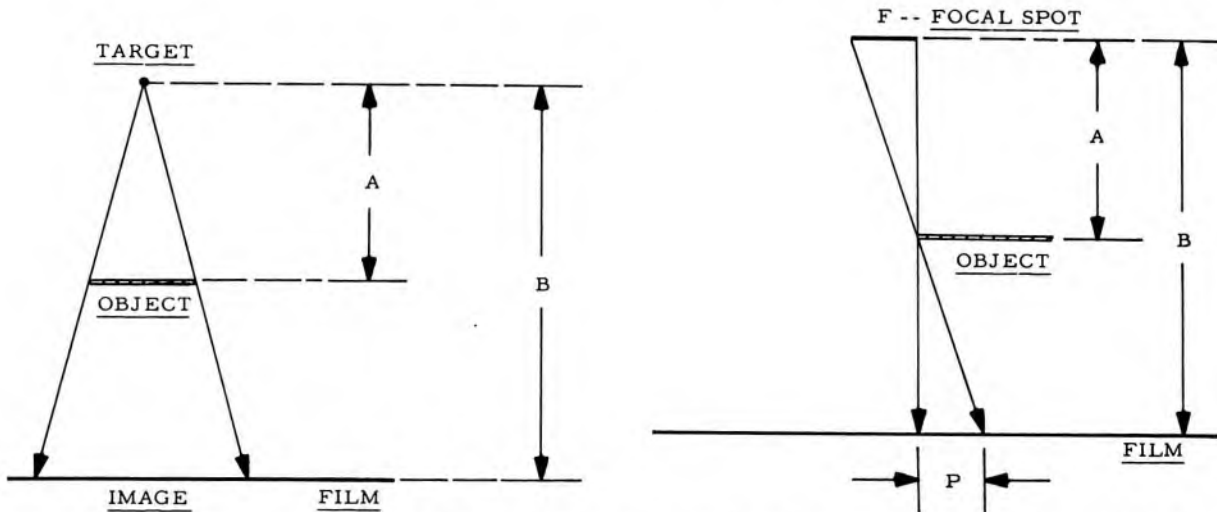


Fig. 20. Geometric relationship in magnification technique. Top, left. X-ray magnification— M . $M = B/A = \frac{\text{Target-film distance}}{\text{Target-object distance}}$. Top, right. Penumbra— P , vs. magnification M . $P = [(B-A)/A] F$, so $P/F = (M-1)$.

Enlargement Technique

Visual images are often enlarged for various effects, as for example in ordinary photography and photomicrography. X-rays

TABLE 8. EQUIVALENT PENUMBRA* VERSUS MAGNIFICATION AND X-RAY TUBE FOCAL SPOT SIZE

Magnification $M = \frac{\text{TFD}}{\text{TOD}}$	Equivalent Penumbra in mm for Focal Spot $F =$				
	0.3 mm	0.5 mm	1.0 mm	1.5 mm	2.0 mm
1.1 [†]	.03	.05	.09	.14	.18
1.2 [†]	.05	.08	.17	.25	.33
1.5	.10	.17	.33	.50	.67
2.0	.15	.25	.50	.75	1.00
2.5	.18	.30	.60	.90	1.20
3.0	.20	.33	.67	1.00	1.33
3.5	.21	.36	.71	1.07	1.43
4.0	.23	.38	.75	1.13	1.50
4.5	.23	.39	.78	1.17	1.60
5.0	.24	.40	.80	1.20	1.60
6.0	.25	.43	.83	1.25	1.67

* This is the ratio of $P/M = F(M - 1)/M$. It indicates the actual blurring effect of penumbra on the magnified image.

[†] 1.1 is typical of mammography and extremities, 1.2 of most other examinations.

Notes: 1. If 0.2 mm equivalent penumbra is acceptable, a 0.3 mm focal spot gives satisfactory results with magnifications of 3.0 times (solid line).

2. If 0.25 mm is acceptable, a 6.0 magnification is permissible (dashed line).

cannot be refracted or simply reflected like light, so x-ray enlargement is accomplished by using divergence of x-rays from a small source. It is evident in Figure 20(Top, left) that each dimension of image O'N' is B/A times larger than the object itself ON. This ratio is called the magnification M:

$$\text{Magnification } M = \frac{\text{Target-film distance TFD}}{\text{Target-object distance TOD}} = B/A \quad (8-5)$$

One may thus enlarge x-ray images by simply increasing the TFD relative to the TOD. There is always an incidental gain in scatter control resulting from the increased air gap.

Unfortunately, the penumbra (P) also increases with magnification. From Figure 20(Top, right), it is seen that penumbra is related to the focal spot size F and magnification as follows:

$$P = F(B - A)/A = F(M - 1) \quad (8-6)$$

Thus, penumbra rises more rapidly than magnification. Consequently, the benefit of image enlargement is ordinarily more than offset by increased blur, and useful magnification procedures require the use of very fine focus x-ray tubes.

It is useful to be able to conveniently estimate how much magnification is permissible for a given focus tube without undue sacrifice of image quality. Evidently the quantity P/M rather than P is of most interest in this regard, since both the image and penumbra size increase with magnification. We shall call P/M the "equivalent penumbra" of the magnified image. From (8-6) we may write:

$$P/M = F(M - 1)/M \quad (8-7)$$

Table 8 shows computed values of P/M for various magnifications and focal spot sizes. Magnifications of 1.2 are typical for most ordinary x-ray studies, and 1.1 corresponds fairly well to usual studies of the breast and extremities.

When excellent detail resolution is required P/M values must be 0.2 mm and less. (Examples include mammography and studies of extremities, foreign body localizations, etc.) It is seen from Table 8 that a 1.0 mm focus permits only 1.2, and even a 0.5 mm focus only 1.67 magnification for this limit of P/M; on the other hand, a 0.3 mm focal spot may be used satisfactorily with a full *three-fold* magnification. For an allowed 0.25 mm value of P/M, the corresponding permissible magnification values are somewhat greater: 1.33, 2.0, and 6.0, respectively.

Unfortunately, the use of small focal spots limits the permissible tube loading considerably. [See Chap. 3, Table 3(A), p. 98.] For example, a 0.3 mm focal spot tube is limited to currents below 50 mA in ordinary designs; this has recently been raised to about 100 mA by use of high speed anode rotation and newer tube designs. The new tube ratings are still relatively low, so magnification techniques have heretofore been limited primarily to applications where patient motion does not limit the final resolution excessively. Where motion blur problems can be adequately controlled, and higher kV techniques are permissible, magnification of the image can overcome limitations in detector resolution by the magnification factor, with some scatter control from the incidental air-gap.

Image intensifiers can respond to relatively low intensity beams. Their use in magnification procedures may greatly expand the use of such techniques.

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9

Basic Principles of Radioactivity

Except for very general discussions in Chapter 1, our coverage has been so far limited to x-rays. Radioactive materials are, however, also widely employed in medicine as well as many other fields. Their clinical application is usually designated “nuclear medicine” and constitutes a useful, rapidly growing medical sub-specialty. In addition to radioactivity, two other nuclear reactions are of great special importance in peace and war: fission and fusion.

In this chapter we discuss all these nuclear reactions, with our main emphasis, however, on radioactivity, its uses, processes, sources, and decay. The following three chapters more specifically discuss applications of radioactivity: basic detection and measurement, clinical measurement techniques, and radiotherapy.

HISTORICAL BACKGROUND

In 1896 A. H. Becquerel discovered that penetrating rays emanate from all uranium salts. These rays readily darken a carefully wrapped photographic plate; further, if one waits somewhat longer, they do so even after traversing fairly thick opaque objects.

This exciting discovery stimulated great investigative efforts, and within two years Marie and Pierre Curie reported the discovery of the radioactive elements polonium and radium. Clinical application of radium followed promptly, as with x-rays after their discovery.

In 1906 Geiger and Rutherford reported a new discharge tube for detecting alpha rays; Geiger and Mueller had improved this tube by 1928, so it could detect other radioactivity rays as well. Since gas ionization and fluorescence were well-known before 1900, many basic scientific principles used in modern radiation measurement existed many years ago.

Other naturally radioactive materials were soon discovered, but they were generally not suitable for diagnostic or therapeutic use. In

1933, however, E. O. Lawrence and M. S. Livingston developed the "cyclotron," and the production of man-made radionuclides began. Cyclotrons and similar particle accelerators were used almost at once to produce small quantities of radioactivity, but the cost was prohibitive and applications were limited to research. (See Chapter 16 for a discussion of particle accelerators and nuclear reactors.)

The Nuclear Age

December 2, 1942, ushered in the large-scale use of nuclear energy. On that historic date Enrico Fermi and his group achieved the first controlled nuclear reaction. During World War II and in the years following there were great military pressures to develop atomic bombs; great advances in nuclear reactor design and construction resulted since the explosive plutonium is made in nuclear reactors. A further impetus has been the increasing operation of reactors for generating electricity. Large quantities of radioactive materials useful in medicine are produced as a by-product of nuclear reactor operation.

In 1946 the newly created United States Atomic Energy Commission (AEC) initiated the medical radionuclide program, which has since expanded steadily and rapidly. More recently many reactor- and accelerator-produced radionuclides have become available, through AEC-licensed pharmaceutical firms, in form appropriate for medical use: presterilized, assayed, and in useful chemical combinations and particle size.

NUCLEAR REACTIONS

Until 1896 only relatively superficial changes in matter were recognized. These affect primarily the relationships *between* atoms and involve at most reactions among their orbital electrons (chemical bonds); however, their atomic nuclei are normally inviolate.

Radioactivity involves *nuclear* upheavals, affecting the character of the atoms themselves. Two other such violent reactions are also of great importance: nuclear *fission* and *fusion*.

In all these nuclear reactions millions of times greater amounts of energy are released per atom than in chemical reactions. Furthermore, this energy results from actual destruction of matter in the nucleus—as much as 0.1 percent of the total! Recall the equation $E = mc^2$, relating the transformation of energy into mass or vice versa. Here energy (E) is in ergs, mass (m) is in grams, and c is the velocity of light in cm/sec. (See discussion of pair production and

annihilation in Chapter 4). The energy appears as gamma rays and the kinetic energy of nuclear fragments and other particles.

We shall now discuss these three processes and their applications.

Radioactivity

Ordinarily atomic nuclei are relatively stable. Thus the ^{16}O and ^1H atoms on the earth's surface show no detectable tendency to change spontaneously into anything else. Some nuclear species, however, *do* show such a tendency; these are called radioactive nuclides or radionuclides. More formally, radioactivity is a process in which nuclei of certain nuclides spontaneously disintegrate, producing new nuclei (normally different chemically), with the ejection of alpha, beta, and/or gamma rays.^{1, 2} The process is inexorable and proceeds independently of any chemical combinations or other variations in the environment of the atoms involved.

EXAMPLES.



When alpha or beta rays are emitted the atom changes in at least two ways. First, Z is usually altered, so the new atom is chemically different. This involves new valence behavior, so previous molecular combinations are generally disrupted. Second, since the nucleus has lost positive or negative charges, the new atom is left charged. In β^- decay, the new atom is neutralized by acquiring an external orbital electron, with the release of the atom's characteristic photon energy. In alpha and β^+ decay, outer orbital electrons are released by the atom to match the loss in positive charge.

Alpha, beta, and gamma rays normally have substantial energy and are hence able to ionize many other atoms in collisions. For example, the average beta ray from ^{131}I has 188 keV of energy. Since on the average about 34 eV are required to ionize an atom in water, just *one* beta ray from ^{131}I potentially can ionize roughly 5,000 atoms in tissue!

Radionuclides are extensively used in clinical diagnostic tests called "tracer" studies. In addition, their ionizing radiation is widely used for the treatment of malignant and some benign conditions. Both

beta and gamma rays are employed clinically, but *not* alpha rays. Materials that emit alpha rays are very hazardous when inside the body because such materials are usually selectively deposited in bone, and alpha rays have great LET values, as shown in Chapter 5.

Table 1 summarizes properties of several commonly employed radionuclides and shows some of their clinical applications.^{3, 4} We shall now briefly consider these applications.

TRACER STUDIES. Much can be learned about bodily processes by following the fate of biologically important chemicals. Studies have been carried out not only in humans, but also in other animals, plants, and microorganisms. Useful studies include: where the material goes and how it is distributed in various organs, tissues, and cells; the routes taken; and the time schedule. In addition, radioactive samples can often be better studied to establish biochemical composition and other information. They can be taken from excreta, blood and other body fluids, surgical specimens, exhaled gases, etc.

Previous to the use of radioactive tracers, studies were carried out using ingenious chemical procedures as well as nonradioactive tracers like dyes. Radioactive tracers, however, conveniently and relatively harmlessly yield data regarding their location *in vivo*, with external measurements. Also, distribution of tagged compounds in organs and even cells is often readily obtained with superimposed microscopic morphology (autoradiographic technique). Finally, extremely tiny quantities of tracer radioactivity are readily measured in samples taken for study. The combination of newer chemical separation procedures and radioactive tracer measurement adds a new dimension to biochemical studies. Chapter 11 discusses tracer measurements in more detail.

ESSENTIAL CHARACTERISTICS OF RADIOACTIVE TRACERS. Two basic characteristics are profitably applied whenever we use radioactive tracers. First, radioactive nuclides behave chemically in the same manner as their stable isotopes. This similarity is in practice quite close for all but certain tritium-carbon chemical bonds, where ³H behavior differs somewhat from that of much lighter ordinary hydrogen. Second, radioactive substances signal their presence by emitting beta and/or gamma rays. (For tracer work gamma emitters are most convenient for sample work and are almost universally required for *in vivo* studies). In addition the rate at which the radioactivity rays are emitted is proportional by definition to the radioactivity and hence the amount of radionuclide at the given location. Valid quantitative measurements of amount and distribution of radioactive tracers are thus made relatively easily.

TABLE 1. SOME DIAGNOSTIC USES OF SEVERAL RADIONUCLIDES

Radio-nuclide	Half-Life	Major Emissions (MeV)		Typical Diagnostic Use and Chemical Form
		Beta	Gamma*	
^3H	12.3 yr	0.018	—	Body H_2O volume, metabolic studies.
^{14}C	5,560 yr	0.155	—	Red cell life span (cohort labeling), metabolic studies.
^{32}P	14.3 d	1.70	—	Red cell survival (DFP), eye and cranial tests at surgery (PO_4^{3-}).
^{51}Cr	27.8 d	—	(EC-9%) 0.32	CrO_4^{2-} : red cell survival, G-I bleeding and protein leakage.
^{59}Fe	45.1 d	0.27 & 0.46	1.1 & 1.29	Ferrokinetic studies; with ^{51}Cr , erythropoiesis and red cell survival.
^{60}Co	5.24 yr	0.306	1.17 & 1.33	Vitamin B12 studies, pernicious anemia.
^{57}Co	270 d	EC	0.12 & 0.14	Same but gives lower whole-body dosage.
^{75}Se	127 d	—	EC & IC 0.121 to 0.402	Pancreas and parathyroid scanning as selenomethionine.
^{85}Kr	10.3 yr	0.67	0.52	L-R shunts.
^{85}Sr	64 d	—	EC-0.513	Bone lesion localization (Sr^{++}).
$^{99\text{m}}\text{Tc}$	6 hr	—	0.140	TcO_4^- : thyroid, brain scanning, albumin tags, blood pool scanning.
^{131}I	8.1 d	0.61	0.364	1. Thyroid as inorg. I: uptake, scan, etc.
^{125}I	60 d	—	EC-0.035	2. Tagged albumin: circulatory studies like blood vol., cardiac output, etc.; placental localization. 3. Rose bengal: liver function. 4. Hippuran: renal function, scan. 5. MAA: lung circulation scanning.
^{198}Au	2.70 d	0.96	0.412	As colloid: pickup by liver, spleen, and other RE cells.
^{203}Hg	47 d	0.21	IC-0.279	1. BMHP: splenic scanning.
^{197}Hg	65 hr	—	EC & IC 0.0776	2. Chlormerodrin: brain and kidney scanning, renography.

*IC means "internal conversion." Instead of gamma emission, orbital electron may be given energy otherwise appearing as a photon, and IC electrons are emitted from the atom.

THREE VARIABLES. Tracer studies can be carried out with much versatility in design and execution. One can vary the material itself, its mode of administration, and the method of measurement.

The material can be varied both as to the radionuclide and chemical form. Generally the radionuclide is chosen on the basis of measurement ease, half-life and particle energies (which also affect patient dosage), and tagging properties. If scanning of reasonably superficial organs is contemplated, photon energy values around 100 keV are preferred because they lend themselves best to collimation and permit use of "gamma cameras." With the development of "radioactivity cows" short-lived pure gamma emitters are now available. These are extremely promising for scanning because large amounts of activity may be used safely. (See Chapter 11.) The best chemical form is crucially related to the study planned and can be quite varied for the same radionuclide.

The mode of administration also depends on the study. When variability of gastrointestinal absorption is now troublesome, oral administration is adequate. Direct intravenous and sometimes catheterized arterial injection is preferred for other work. In most cases true solutions are employed. Colloidal ^{198}Au can be used, however, to study liver and spleen, and macro-aggregate ^{131}I serum albumin gives graphic scans demonstrating blood circulation in the lung.

Measurements are made both in vivo and by means of samples. In vivo studies can be performed locally, or a regional area can be scanned. The scan gives much more information but at present takes more time and (normally) more radioactivity for good results. One can, in addition, take sequential measurements of radioactivity of an organ to estimate its accumulation and clearance functions. Other useful information is often obtained from samples such as urine and feces; blood, saliva, and other body fluids; surgical specimens; exhaled gases; etc.

THERAPY (TABLE 2).⁵ The beta and gamma rays of radionuclides ionize and excite atoms in a manner similar to x-rays and hence produce similar tissue reactions. Beta rays and primary electrons from x-rays have sufficiently comparable LET values that their biologic effects are generally similar.

Radionuclides are employed therapeutically both in sealed form and as solutions.

The most intense sealed sources are those ranging from 100 to 10,000 curies or more of cobalt-60 and cesium-137 used in teletherapy machines. These radionuclides emit gamma rays, which are used for external beam therapy instead of x-rays. Such enormous activity sources are housed in special well-shielded containers with appro-

TABLE 2. THERAPY APPLICATIONS OF RADIONUCLIDES⁴

Radio-nuclide	Half-Life	Major Rays		Typical Chemical Form and Use
		Beta (MeV)	Gamma (MeV)	
¹³¹ I	8.1 d	0.61	0.364	As inorganic iodine: hyperthyroidism; anginal relief in euthyroids; thyroid cancer
³² P	14.3 d	1.70	—	As soluble PO ₄ ³⁻ , polycythemia vera, chronic leukemia; as CrPO ₄ , ascites relief.
¹⁹⁸ Au	2.70 d	0.96	0.412	As colloid, ascites relief; experimental preoperative use in breast cancer.
B. As Metals, Local Application (Short Half-Life Nuclides)				
⁹⁰ Sr & ⁹⁰ Y	27.7 yr	0.54 & 2.26	—	Eye application, superficial benign lesions.
⁹⁰ Y	64 hr	2.26	—	Pellets and wires, local irradiation (surgical insertion).
¹⁹² Ir	74.4 d	—	0.283 to 0.613	Seeds in nylon tubing, interstitially (vs. Ra needles).
C. Gamma Emitters of Long Half-Life				
⁶⁰ Co	5.24 yr	0.306	1.17 & 1.33	As metal 1 cm to 1 inch dia. cylindrical sources, up to over 10,000 <i>curies</i> in teletherapy. Also, less frequently as metal in needle and capsule form, as Ra substitute.
¹³⁷ Cs	26.6 yr	0.52	0.662	Larger and less intense teletherapy sources, as Cs salt. <i>Also as ceramic salt</i> , more recently as radium substitute in needles and capsules.
²²⁶ Ra	1,622 yr	—	wide range 0.24 to 2.20	As radium salt in radium capsules and needles, for use adjacent to and in tumor tissue.

privately designed shutters and adjustable collimators. (These units are discussed in Chapter 16.)

Radium sources and their substitutes are used in close contact with malignant tissue, alongside the lesion (capsules) or interstitially (needles). Due to their proximity to the lesion, minimum dosage is delivered from such sources to surrounding healthy tissues.

Some small superficial eye lesions have been treated using the beta rays from strontium-90 radioactive sources. Commercial applicators employ a very thin aluminum cover which transmits the beta rays but confines the radioactive material.

Both true and colloidal solutions of many radioisotopes are employed therapeutically. True solutions are administered in a form selectively concentrated by the target organ, which is thus selectively treated by concentrating the radioisotope. Examples are inorganic ^{131}I used to treat thyroid carcinoma and hyperthyroidism, as well as to induce near-myxedema for relief of angina; and phosphorus-32 in phosphate form used to treat polycythemia vera and chronic leukemia. It must be emphasized that other tissues (hemopoietic, renal, etc.) are inevitably also treated at the same time. Colloidal solutions or suspensions like gold-198 and phosphorus-32 as chromic phosphate have been widely used in the pleural and peritoneal cavities to reduce ascitic fluid formation.

Nuclear Fission

Under certain circumstances nuclei of very heavy elements (Z greater than 89) split into nearly equal parts and release neutrons and other radiation. This process is referred to as "fission."⁶ The product atoms generally have Z values of 34 to 57. Unlike radioactivity, fission almost always requires the intrusion of a neutron. (This is true of almost all such elements. ^{254}Cf is a man-made radionuclide which undergoes *spontaneous* fission with a half-life of 54 days.) Moreover, a neutron that is too fast or too slow has a very limited likelihood of interaction. The most important practical fissionable materials are ^{235}U and ^{239}Pu ; in addition, ^{238}U and several other isotopes can undergo fission under suitable circumstances.

Cosmic radiation provides neutrons to initiate fission of heavy atoms. Natural fission is rare, however, because fissionable atoms are too dispersed to maintain a self-sustaining fission process. Before large-scale fission can occur large numbers of atoms must participate. This can be accomplished in specially engineered bombs and reactors, utilizing the principle of the "chain reaction." The atomic bomb employs fission in an uncontrolled reaction. Nuclear reactors employ fairly elaborate control systems to assure a controlled release of energy

and fission products and may be used to produce electric power as well as useful radioactive isotopes. These points are discussed more fully in Chapter 16.

Nuclear Fusion

Nuclei normally repel each other since they are all positively charged. For distances down to considerably less than 1 Å the force of repulsion apparently obeys the inverse square law. However, at closer separations approximating nuclear dimensions new mysterious forces take over, causing some nuclei to fuse together.⁷ These forces are undoubtedly closely related to nuclear binding forces generally, which keep protons confined in nuclei despite their mutual electrical repulsion.

One should theoretically be able to fuse some small nuclei together to make bigger ones, if only he can provide enough energy to bring them close enough together. Fusion of very light nuclei can release tremendous energies because the new nuclei have less mass than existed in the original particles. The lost mass is all converted into energy. Thus, if one could coax such nuclei to fuse he would have a source of enormous energy. The repulsion forces are least in light nuclei because they have fewer protons, and hydrogen is the likeliest fusion material.

Actually the sun and other stars routinely accomplish fusion, coalescing hydrogen nuclei into those of helium and probably other elements as well. This is most important to us since life on earth could not exist without the sun's energy. Fusion is possible in the interior of stars like our sun because they provide two essential ingredients for a fusion furnace. First, great gravitational fields confine the fuel like an oven, so it can fuse efficiently. Second, temperatures above 20 million degrees absolute are readily maintained. At such temperatures nuclei acquire tremendous velocities, enough to overcome their mutual electric repulsion.

The hydrogen bomb is in principle a small man-made sun. The fuel is usually heavier isotopes of hydrogen in the compact form of alkali hydrides like Li^2H . High temperatures are achieved by using an ordinary fission or atomic bomb. A suitable container confines the fuel long enough to insure a reasonably complete reaction. The products are helium and neutrons.

If controlled, fusion could provide virtually unlimited power, using plentiful ^2H as fuel. Much research is in progress in the fields of magneto-hydrodynamics and similar new disciplines. Such efforts, if successful, could help in the agricultural and industrial expansion of many areas presently limited in energy resources. An advantage of

a *fusion* over the *fission* reactor is that its products are nonradioactive. In fission there is inevitable production of ^{137}Cs , ^{90}Sr , and other long-lived fragments of the fissioning fuel atoms. These present problems of dangerous waste disposal which may ultimately set practical limits on the expansion of fission reactor installations.

RADIOACTIVITY—PROCESSES AND SOURCES

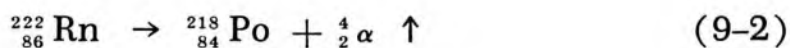
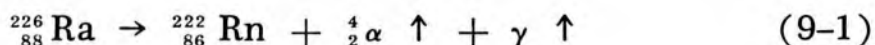
Of all nuclear reactions, radioactivity is the only one so far directly applicable to clinical work. In this section we describe radioactivity processes and how radionuclides are obtained.^{8, 9}

Radioactive Decay Processes

Five decay mechanisms are of major interest here: α , β^- , β^+ , electron capture, and isometric transitions.¹⁰

ALPHA DECAY. Alpha particles are ejected with enormous energies (2 to 8 MeV) from nuclei of heavy elements, of atomic numbers 84 and higher. They are helium nuclei (2 protons and 2 neutrons) and hence have twice as much charge and about 7,300 times as much mass as electrons. Because of their greater mass they travel much more slowly than electrons of the same kinetic energy. Consequently, they interact readily with atoms in tissue, yielding high LET values (Fig. 4 in Chap. 5). The net result is that their penetration is *low* and their *RBE* high.

Release of an alpha particle removes four nucleons from the nucleus, two of them protons; hence *A* is reduced by 4, *Z* by 2. For example:



Note that gamma rays are also produced in (9-1). (Beams of alpha, beta, or neutron particles are often loosely called "rays.") Generally, gamma rays may be released in both alpha and beta decay when the new nucleus is left with excess energy. Recall that nucleons are arranged in different energy levels within the nucleus, analogous to electrons in their orbits. When nucleons drop from a higher to a lower energy level during return of a nucleus from an excited state, gamma rays are released similarly to characteristic x-rays. *Thus gamma rays are essentially nuclear characteristic rays.* Sometimes the

nucleons in the new nucleus are already at their steady state levels. No gamma rays are then released, as in decay of ^{222}Rn , ^{14}C , ^{32}P , etc.

Alpha particles are emitted with characteristic energies, analogous to characteristic x-rays. For example, *all* alpha rays from decay of ^{222}Rn have identical energy (6.28 MeV); those of ^{226}Ra are in *two discrete groups*, 4.79 and 4.61 MeV, while those from reactor and bomb fuel plutonium (^{239}Pu) are in *three groups*, 5.096, 5.134, and 5.247 MeV. Alpha particles of a radionuclide thus have characteristic energies and penetrate characteristic distances in soft tissue.

In any radioactive decay process a fixed total amount of energy is released with each disintegration. There are, however, a great many nucleons in the nucleus. Depending on which particular one or ones participate, the new nucleus may be left more or less excited. When ^{222}Rn decays, the new nucleus is *never* left excited, and no gamma rays are emitted. In ^{226}Ra decay, this is also true in general; however, in 1.2 percent of disintegrations, a weak gamma ray is emitted. Figure 1 shows so-called "decay scheme diagrams" which conveniently summarize such information, for ^{222}Rn and ^{226}Ra .

BETA DECAY. Beta particles are either minus or plus electrons and designated β^- and β^+ . The nucleus contains no electrons, only neutrons and protons, so one might ask where these electrons come from. It is generally accepted that they are produced in nuclei by transformation of neutrons into protons and vice versa. Minus beta particles

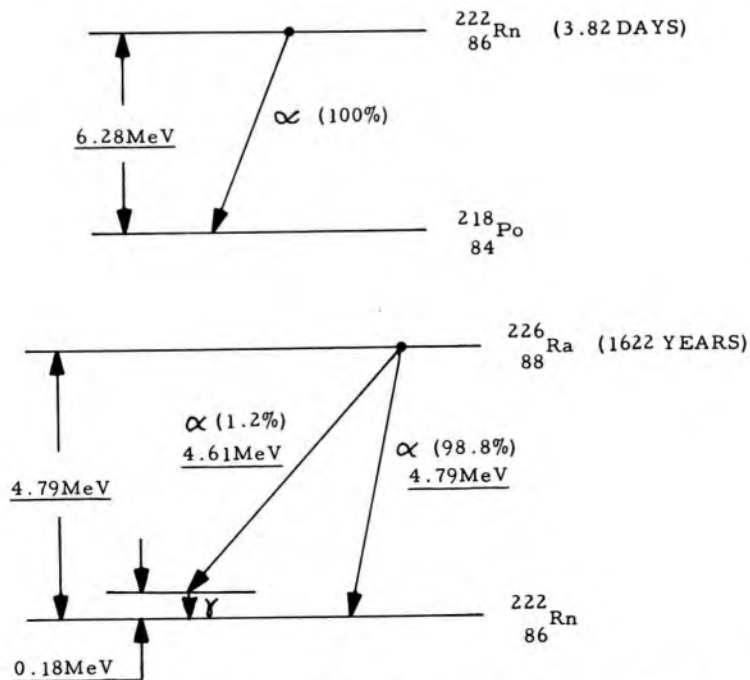
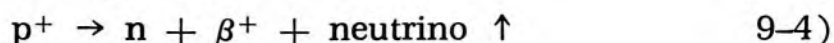
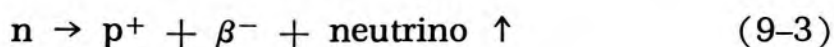


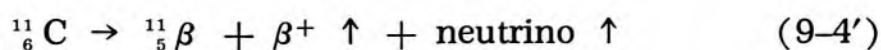
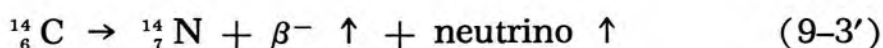
Fig. 1. Alpha decay: examples. Top. Radon-222 \rightarrow polonium-218. Bottom. Radium-226 \rightarrow radon-222.

(β^-) are identical to orbital electrons; plus beta particles (β^+) are *positrons*, identical to those produced during pair and triplet production.

β^- decay occurs in *neutron-rich*, β^+ decay in *proton-rich* nuclei. The basic *descriptive* reactions are as follows:

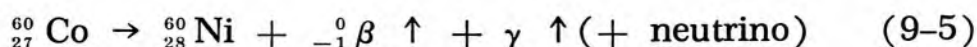


There are apparently two different kinds of neutrinos, ordinary and antineutrinos. That emitted in β^- decay is actually an antineutrino. We simply note but do not stress the distinction. These reactions refer to events in radioactive nuclei only, with many other nucleons present. Note there is no change in A , but Z is increased and decreased in β^- and β^+ decay, respectively. For example:



Since beta particles are charged and leave the nucleus, it is clear Z must change accordingly. The atomic mass number remains the same because the *total number* of nucleons is unchanged, only the relative number of protons and neutrons.

As in alpha decay, gamma rays may also be produced by beta decay if newly formed nuclei are left in an excited state. For example:



Energy diagrams may be drawn, similar to these of alpha emitters. Figure 2 presents some typical examples of beta decay and of other reactions to be described below.

BETA RAY CONTINUOUS SPECTRA. Neutrinos are neutral particles which are extremely light compared with electrons. Their existence helps explain the *continuous* spectrum of beta ray energy, which differs from that of alpha and gamma rays. It will be recalled that alpha and gamma rays from a given radionuclide have discrete energies (i.e., discontinuous, characteristic spectra). This is reasonable to expect, since radioactive disintegration involves reassembly of nucleons into a lower energy configuration with the release of a unique and fixed amount of energy for each rearrangement. In alpha or gamma emission, *all* the energy available to each emitted particle is

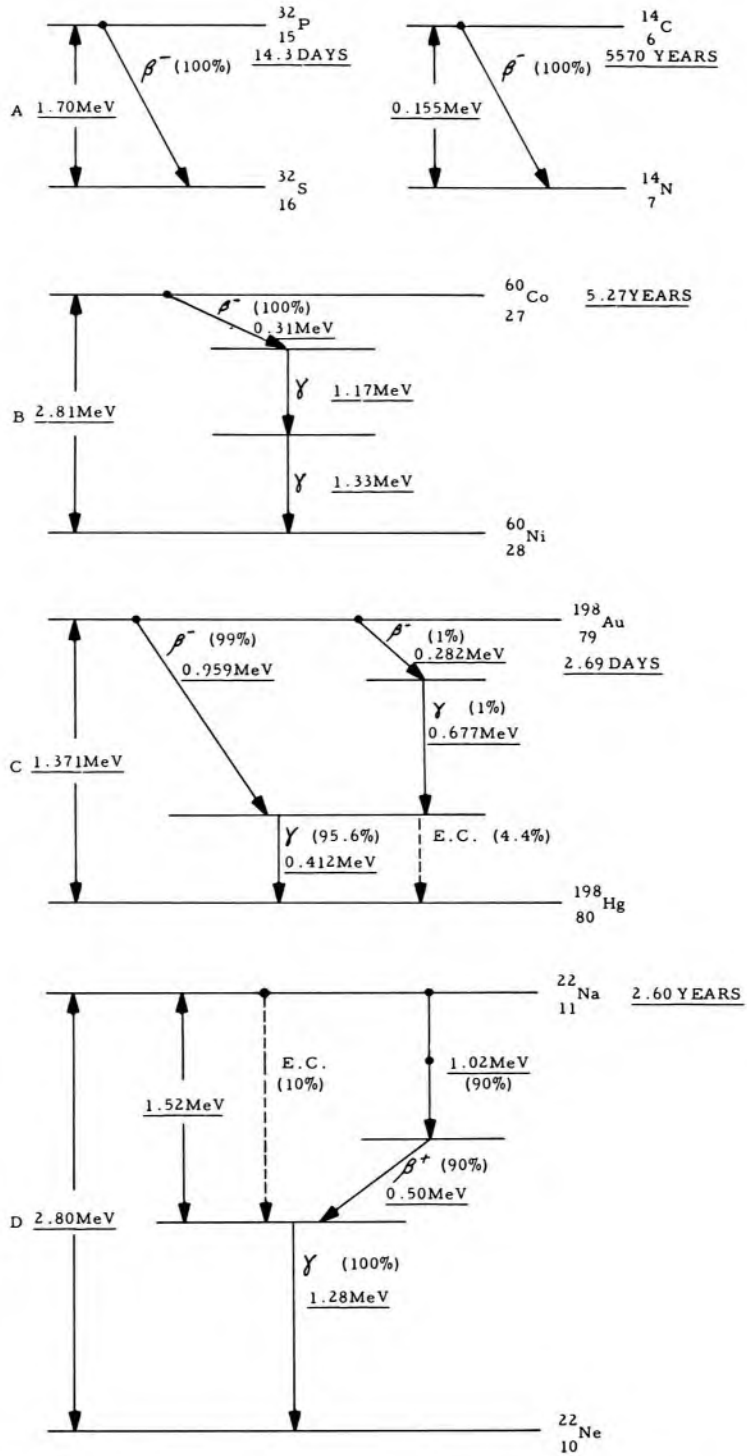


Fig. 2. Beta decay: four examples. A. Simple β^- decay—phosphorus-32 and carbon-14. B. β^- plus gamma decay—cobalt-60. C. β^- plus gamma decay, more complex, with electron capture also—gold-198. D. β^+ plus gamma decay, also more complex, with electron capture—sodium-22. Note: 1.02 MeV of available nuclear energy is used in producing β^+ , so it emerges with this much less than the 1.52 MeV available to the total disintegration.

delivered to it completely. Neither ever shares its energy with a neutrino.

In beta decay the situation differs basically. Nucleons actually change their *nature* rather than simply their *location*. In the process of this more complex transformation not only beta rays but also neutrinos are emitted. These leave with some of the fixed amount of energy available to the beta transformation, so the beta ray emerges with only *part* of the energy. The exact neutrino energy ranges from zero (when the beta ray receives all the available energy) to substantially the entire available energy (leaving negligible energy for the beta ray). The existence of neutrinos has been confirmed in other work, and this hypothesis is generally accepted.

Figure 3 shows the distribution of the beta rays from ^{32}P . E_β , the maximum beta ray energy, is 1.70 MeV; it is evident beta rays

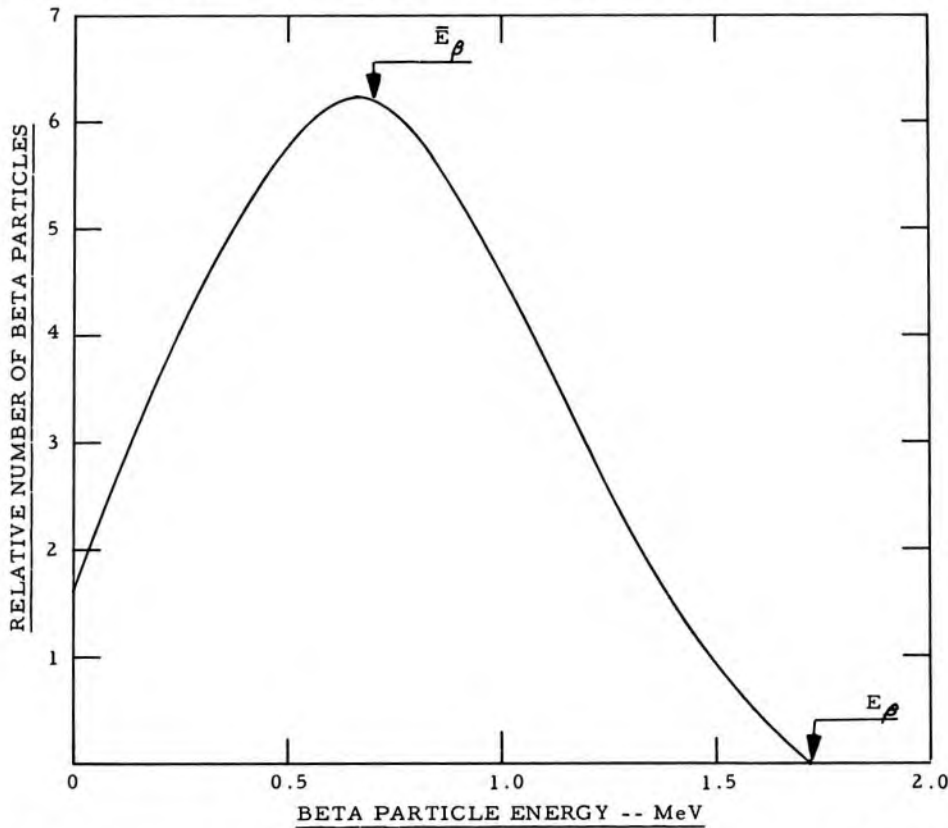


Fig. 3. Continuous spectrum of beta decay of phosphorus-32. Although 1.70 MeV of energy was available for the emitted particles, only a few electrons even approach this maximum value (E_β). The average energy \bar{E}_β is actually 0.70 MeV. The actual beta particle energies range from 1.70 MeV down to as low as zero. The reason is that neutrinos also are emitted during beta decay. They too emerge with energy—from 1.70 MeV down to zero. The energy partition explains the continuous spectrum. Note alpha and gamma energies are always the full value, and their spectra are line, or characteristic, spectra. (Redrawn from Sinclair.¹¹ Courtesy of Charles C Thomas, Publisher.)

TABLE 3. BETA-RAY ENERGY OF SEVERAL CLINICALLY USEFUL RADIONUCLIDES

Radio-nuclide	Energy, MeV		Range in Water or Soft Tissue, mm	
	Maximum (E_{β})	Average (\bar{E}_{β})	Maximum	Average
^{90}Y	2.18	0.93	10.5	3.65
^{32}P	1.70	0.70	7.8	2.5
^{198}Au	0.96 (99%)	0.328	3.8	0.9
	1.37 (1%)			
^{85}Kr	0.67 (99.3%)	0.22	2.4	0.48
	0.15 (0.7%)			
^{137}Cs	0.52 (92%)	0.23	1.7	0.45
	1.17 (8%)			
^{131}I	0.61 (87%)	0.188	2.1	0.37
	0.36 (9%)			
	0.25 (3%)			
^{59}Fe	0.46 (53%)	0.118	1.45	0.18
	0.27 (46%)			
^{60}Co	0.306	0.094	0.78	0.12
^{14}C	0.155	0.050	0.27	0.043
^3H	0.018	0.006	0.008	0.0015

Notes:

1. Radionuclides listed in order of descending beta particle range.
2. Ranges obtained from corresponding beta-ray energies, using Quimby's data.¹²

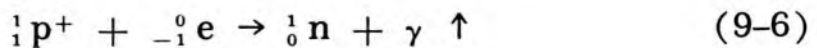
are produced with energies down to practically zero, with an average beta energy (\bar{E}_{β}) of about 0.70 MeV.

What is the practical significance of the continuous spectrum of beta rays? Primarily it reduces their penetration in tissue. The *maximum* penetration of beta rays is determined by the maximum energy (E_{β}). Generally, however, the rays penetrate far less in accordance with their average energy (\bar{E}_{β}). Table 3 shows values of E_{β} and \bar{E}_{β} for several clinically useful beta emitters and also the resulting estimated maximum and average penetration of these rays in soft tissue.¹² Note that beta rays travel from 0.1 mm up to over 10 mm in tissue, maximum; average distances are less by factors of about 3 to 6 times. In view of the limited average beta range of most radioisotopes, it is evidently fruitless to attempt definitive cancer therapy with beta rays when they must traverse much more than a millimeter of water or tissue to reach viable tumor cells.

FATE OF BETA PARTICLES. β^- rays interact with atoms in just the same way as any other fast electrons. Although bremsstrahlen production can of course occur if they strike nuclei, this is ordinarily not very likely in tissue, in which atomic numbers are quite low. (For example, only 0.5 percent of ^{32}P beta ray collisions produce bremsstrahlen in tissue. Significant bremsstrahlen, however, may be produced by ^{32}P and ^{90}Sr stored in therapeutic amounts in high Z applicators or bottles.) Hence, beta rays traversing tissue expend virtually all their energy ionizing and exciting atoms. Generally speaking, $\bar{\beta}$ ray energies are somewhat greater than those of recoil electrons; however, apparently no clinically significant difference in biologic effectiveness results from this fact. The β^- ray ends up a free electron similar to other such electrons normally present in matter.

What of β^+ rays? Initially these interact with atoms quite similarly to β^- rays of the same energy. As described previously (Chap. 4), however, a positron or β^+ particle is an unfortunate stranger in our universe; at sufficient low speed it combines with an ordinary electron, bringing about their mutual destruction. The product of the sad union is a pair of annihilation photons of 0.51 MeV each, corresponding to the lost "rest" mass of each electron. These photons emerge in opposite directions, a fact potentially useful in precision localization of β^+ emitters in the body, using coincidence counting techniques. These make use of the fact that annihilation photons emerge simultaneously in opposite directions. Two detectors are used, and only signals received simultaneously by both detectors (*coincident*) are detected. Others can then be ignored, so only desired signals are registered. If absorbed in the patient, annihilation photons contribute to his whole body dosage. Because it can release 1.02 MeV by annihilation later the β^+ particle must be supplied with this energy when it is created. This comes from the reacting nucleus [Fig. 2(D)].

ELECTRON CAPTURE (EC). Alpha and beta emissions involve decay processes in which active nuclei simply eject particles to achieve a less energetic state. Some man-made radionuclides of the proton-rich type decay in an alternative way to β^+ emission. Instead of the offending proton's changing into a neutron by losing a β^+ , it does so by capturing an ordinary electron from a nearby orbit:

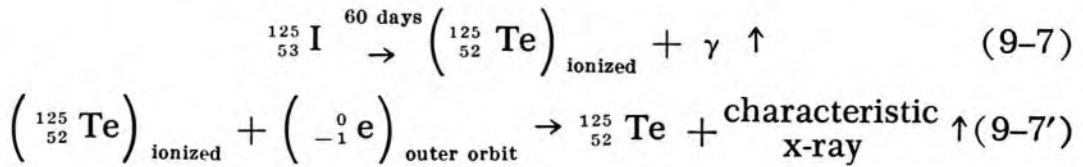


The nearest electron is the most likely candidate for this acquisition, so K-orbit capture is most frequent, although, to a lesser extent, L-orbit and even M- and N-orbit captures sometimes occur.

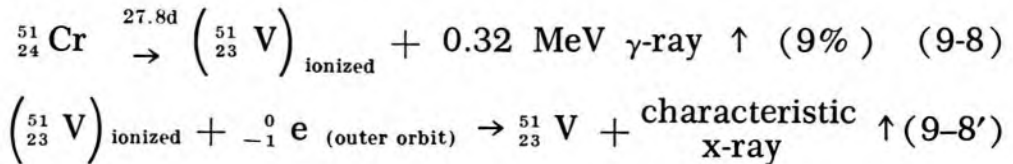
Electron capture is a rather low power affair, and relatively low gamma ray energies usually result (Table 1). In addition, often the gamma ray is not produced at all; apparently only certain protons in the nucleus release energy when neutralized (for example, only 9 percent of those of ^{51}Cr).

Of course, the new atom is always left ionized with a K- or L-orbit location empty. (Also, since Z has been reduced by 1, the outer orbit has an extra electron.) Characteristic x-rays are, therefore, always produced when electron capture occurs.

Two examples may help clarify the process. First consider ^{125}I :



For ^{125}I roughly every fifth electron capture is from the L orbit. Another possibly more useful radioisotope is ^{51}Cr :



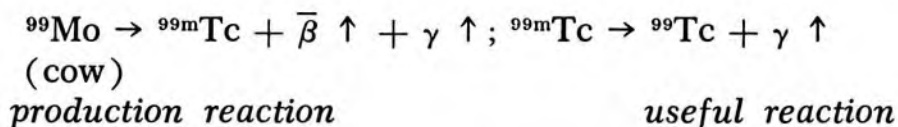
In summary, electron capture involves the use of one of the atom's own K- and sometimes L-orbit electrons to convert one of its protons into a neutron. In at least some reactions a gamma ray is emitted. The new atom is left with an inner orbital vacancy. This is restored in the usual way, with production of characteristic x-rays.

EC radionuclides have a distinct safety advantage in human tracer studies because there are no beta rays. (We shall discuss this more fully in Chapter 12.)

ISOMERIC TRANSITION (IT). It will be recalled that gamma rays may be emitted from the product nuclei after α , β , or EC reactions take place (more generally, following any nucleon alterations). Normally there is no significant delay in the release of the gamma rays. Some product nuclei, however, retain their extra energy for substantial periods. Moreover, they emit their delayed gamma rays according to a time schedule similar in nature to that of radioactive decay ("exponential decay"—see below). If the product nuclei are not themselves radioactive, the reaction provides a pure gamma ray, often very useful for in vivo measurement, without beta dosage to the patient.

The most exciting IT radioisotope at this writing is $^{99\text{m}}\text{Tc}$. This is used as the pertechnetate ion and is produced as needed in the

laboratory from a ^{99}Mo parent or “cow.” (A radioactivity “cow” is a device for obtaining a relatively short-lived radionuclide by “milk- ing” a preparation of a parent, longer-lived radionuclide. The latter is the source of the former. For example,



Its photon energy is almost ideal for scanning, and its short half-life permits its safe use in high activity doses to the patient, further facilitating measurement.

Table 4 summarizes essentials of the five radioactive decay processes.

Sources of Radionuclides

Some radionuclides occur naturally; others are man-made.

NATURAL SOURCES. Most natural radionuclides are simply survivors from those evidently present on earth since its formation. For example, ^{238}U , ^{235}U , and ^{232}Th have half-lives (defined below) of 4.5, 0.71, and 13.9 *billion* years, respectively, indicating only ^{235}U has been substantially depleted in the billions of years of the earth's existence! Each of these decays into new radionuclides, and they in turn decay into others. Ultimately, after many steps, the sequences terminate in the familiar stable isotopes of lead of atomic mass numbers 206, 207, and 208, respectively. As a result, each of the original three radioisotopes produces a large number of “daughter” radioisotopes in three separate sequences, or “series.” For example, ^{226}Ra arises from the decay of ^{238}U after five steps.

Another interesting “old-timer” is ^{40}K . This is present naturally, mixed with ordinary stable ^{39}K and ^{41}K , constituting 0.12 percent by weight. It emits both energetic beta and gamma rays (1.4 and 1.5 MeV, respectively) and decays very slowly, with a half-life of 1.4 billion years. A standard man, however, contains 1/6 gram of ^{40}K : this is a millionth of a microcurie, possibly a genetically significant amount.

Finally, ^{14}C also occurs naturally. This decays quite rapidly in comparison with the above four radionuclides. However, it is continuously produced in air from nitrogen as a result of bombardment by fast neutrons produced by cosmic radiation. A more recent source of fast neutrons has been nuclear bomb explosions, which have already contributed sufficiently to the total atmospheric ^{14}C that future archaeologists may be misled in their attempts at carbon dating!

TABLE 4. RADIOACTIVE DECAY PROCESSES

1. Alpha Decay: nucleus *reduced* in Z by 2, in A by 4.
 - (A) He nucleus emerges, of 2 protons and 2 neutrons.
 - (B) Energies lie between 2 and 8 MeV.
 - (C) Travel 40μ or less in tissue.
 - (D) Characteristic or line spectrum.
 - (E) If new nucleus is left energized, gamma rays also released.
2. Beta (-) Rays: nucleus *increased* in Z by 1, *unchanged* in A .
 - (A) Particle has same mass and charge as ordinary electron. Produced in nucleus by conversion of neutron to proton.
 - (B) Energies usually well below 2 MeV.
 - (C) Range in tissue rarely exceeds 10 mm; usually less than a few mm.
 - (D) Continuous spectrum, with neutrino release.
 - (E) If new nucleus is left energized, gamma rays are also released.
3. Beta (+) Rays: nucleus *decreased* in Z by 1, *unchanged* in A .
 - (A) Particle is similar to positron produced in pair production. Produced in nucleus by conversion of proton to neutron.
 - (B) Continuous spectrum, with neutrino release.
 - (C) If new nucleus is left energized, gamma rays are also released.
 - (D) After conveying most of its energy to the medium, a beta (+) particle combines with an ordinary electron to yield two 0.51 MeV annihilation photons.
4. Electron Capture (EC): nucleus *decreased* in Z by 1, *unchanged* in A .
 - (A) Alternative to beta (+) decay.
K-orbit (or less frequently L-orbit) electron combines with nuclear proton, converting it into a neutron.
 - (B) *Two rays may result, both of relatively low energy.*
 - (1) Gamma ray, from energization of new nucleus. This is not always present, as it depends on *which* proton was changed.
Example: only 9% of ^{51}Cr disintegrations yield gamma rays.
 - (2) Characteristic x-ray: from restoration of K- or L-orbit electron (always 0.1 MeV or less).
5. Isomeric Transition (IT): nucleus *unchanged*.
 - (A) Simply delayed release of gamma ray by a new nucleus.
Corresponds to phosphorescence in characteristic photon release by excited atoms.
 - (B) Spontaneous, with exponential decay like ordinary radioactivity.
 - (C) Produces pure gamma rays. Resulting lowered patient dose makes use of such radioisotopes attractive.

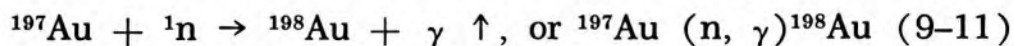
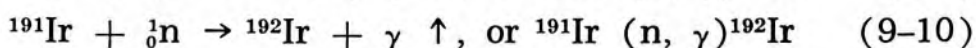
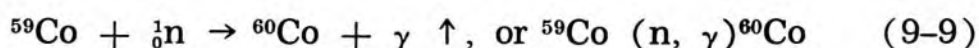
Of all the natural radionuclides ^{226}Ra is the most useful medically. It is employed mainly in the treatment of malignant disease by local application, utilizing gamma rays produced by its daughter products ^{214}Pb and ^{214}Bi , so-called "radium-B" and "radium-C." (See Chapter 12.)

MAN-MADE. Man-made radionuclides are generally produced in nuclear reactors, or cyclotrons and similar particle accelerators. In nuclear reactors fission of heavy nuclei like ^{235}U or ^{239}Pu produces

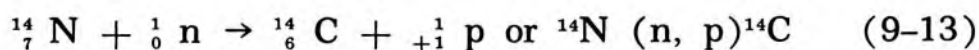
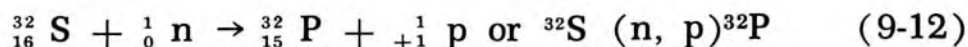
both fragment atoms and neutrons in great abundance. Since they are products of fission of neutron-rich materials, the fission fragment radionuclides are also rich in neutrons, and decay by β^- decay (e.g., ^{137}Cs , ^{90}Sr , ^{131}I , ^{85}Kr , etc.). The neutrons released during fission may be used to interact with nuclei of other atoms, to increase their neutron content (see below). The resulting radionuclides are then also neutron-rich and therefore β^- emitters.

Accelerators are generally used to speed up protons or deuterons. They therefore bombard target materials with protons and tend to increase the proton-neutron ratio in target atoms. The new nuclides tend to be proton-rich and to decay by β^+ or EC processes.

Two types of neutron interaction—slow and fast—can take place in reactors. If slow neutrons are used, suitable irradiated nuclei simply “capture” neutrons. In the process new nuclei are produced, chemically the same as before but heavier by one neutron; in addition a gamma ray is usually released immediately. The reaction is referred to as an (n, γ) reaction (n goes in; γ comes out). For example, the following reactions are used routinely to produce useful radioactive nuclides from the stable nuclide:



If faster neutrons are used, their action is more violent: the incoming neutron can actually drive a proton out of the nucleus, taking its place. The result is a new element because Z is *decreased* by 1. Since the proton is replaced by a neutron, the new nucleus has the same atomic mass. (In the process the atom also loses an electron, since the nucleus has one less positive charge.) This reaction is referred to as an (n, p) reaction, since a neutron enters the nucleus and a proton leaves. For example:



Accelerators are useful for producing β^+ and EC radioisotopes. For example, ^{22}Na is made this way.

A discussion of procurement procedures is given in Chapter 15, but it appears appropriate now to clarify specialized terminology used by suppliers of radionuclides. Radioactivity units are defined in

TABLE 5. UNITS OF RADIOACTIVITY

Unit	Definition	Abbreviation
basic	disintegration per second	d/s
curie	37 <i>billion</i> d/s (3.7×10^{10} d/s)	Ci
millicurie	37 <i>million</i> d/s (3.7×10^7 d/s)	mCi
microcurie	37 <i>thousand</i> d/s (3.7×10^4 d/s)	μ Ci
1 megacurie	1,000,000 curies or 10^6 Ci	MCi
1 kilocurie	1,000 curies or 10^3 Ci	kCi
1 millicurie	.001 curie or 10^{-3} Ci	mCi
1 microcurie	.000001 curie or 10^{-6} Ci	μ Ci
1 nanocurie	.000000001 curie or 10^{-9} Ci	nCi
1 picocurie	.000000000001 curie or 10^{-12} Ci	pCi

TABLE 6. RADIOACTIVITY SHIPMENT TERMINOLOGY

A. Activity

- Total in shipment, normally in microcuries or millicuries.
- Concentration of activity in shipment.
 - Solutions: mCi/ml of solution.
This is usually called the solution "specific activity."
"Specific concentration" would appear to be more appropriate.
 - Solid material: mCi/g or Ci/g of material.
This is also usually referred to as "specific activity."
Examples:
 - ^{60}Co teletherapy source: 26 Ci/g of metal.
 - ^{14}C compound: 500 μ Ci of ^{14}C in 10 g dry compound; specific activity is 50 μ Ci/g.

B. Carrier

- This is stable isotope mixed with desired radionuclide, in the same chemical form.
Example: ^{127}I added to ^{131}I in water solution.
- Purpose: to chemically stabilize some extremely low concentration solutions of radionuclides, which can otherwise deteriorate. Is present as impurity inherent in some production methods.
Example: ^{197}Au with ^{198}Au , ^{59}Co in ^{60}Co , ^{127}I with ^{131}I , etc.

Table 5 and discussed below. Table 6 briefly summarizes widely used terms which can often appear ambiguous or confusing to new users.

RADIOACTIVE DECAY

Our discussion so far has been primarily qualitative, dealing with processes, applications, and sources of radioactivity. Intelligent appli-

cation of radionuclides, however, requires fairly accurate knowledge of the amounts involved. In all applications the degree of radiation hazard increases with the amount handled or administered. In tracer work accurate measurements of activity are required, and calculation of organ and systemic dosage requires knowledge of the amounts and distribution of radioactivity. We shall hence briefly define various terms and concepts essential to understand the quantitative evaluation of radionuclides. These include the "amount" of radioactivity, arithmetic of radioactive decay, and the millicurie-hour concept.

Amount of Radioactivity and Units

In general, when we speak of the amount of any activity, we have in mind the frequency of events. This could be, for example, the number of patients treated per week, the number of people entering a place of business per hour, the number of automobiles crossing a bridge per day, etc. Similarly, the amount of radioactivity, or activity, essentially means the number of events in a given time yielding the emitted rays. Thus, activity is the rate of nuclear disintegrations, and the unit is disintegrations per second (d/s). For example, one might have 10,000 d/s of ^{131}I and 5,000 d/s of ^{32}P ; there is then twice as much ^{131}I as ^{32}P activity.

In practice much larger units are required for convenience, and the *curie* and its submultiples are more useful than d/s (Table 5). The curie, named in honor of the discoverers of radium, Pierre and Marie Curie, was defined essentially as the radioactivity of 1 gram of radium element. Years later this was established to be approximately 37 billion d/s, and the present definition has settled on this figure. The most commonly used units are *curie* (Ci) and *kilocurie* (kCi) for teletherapy, *millicurie* (mCi) for other therapy, and *microcurie* (μCi) for tracer work. Nanocurie (nCi) and picocurie (pCi) units are convenient for dealing with measurements of very tiny quantities of radioactivity. These arise in radiation safety and ecologic studies of natural and man-made radioactivity in living organisms.

Initially an attempt was made to define radioactivity in terms of the mass of the element involved. The activity is what is actually desired, however. Although the activity and mass of a particular radioisotope always increase together, activity is usually much easier to measure. For example, one μCi of ^{131}I is associated with only .000008 μg of ^{131}I , with similarly tiny amounts for other useful radionuclides. Such tiny masses are obviously extremely difficult to measure and handle, whereas 1 μCi of activity is usually conveniently measured. Note that the mass required to yield a given activity depends on the radionuclide involved. (See below.)

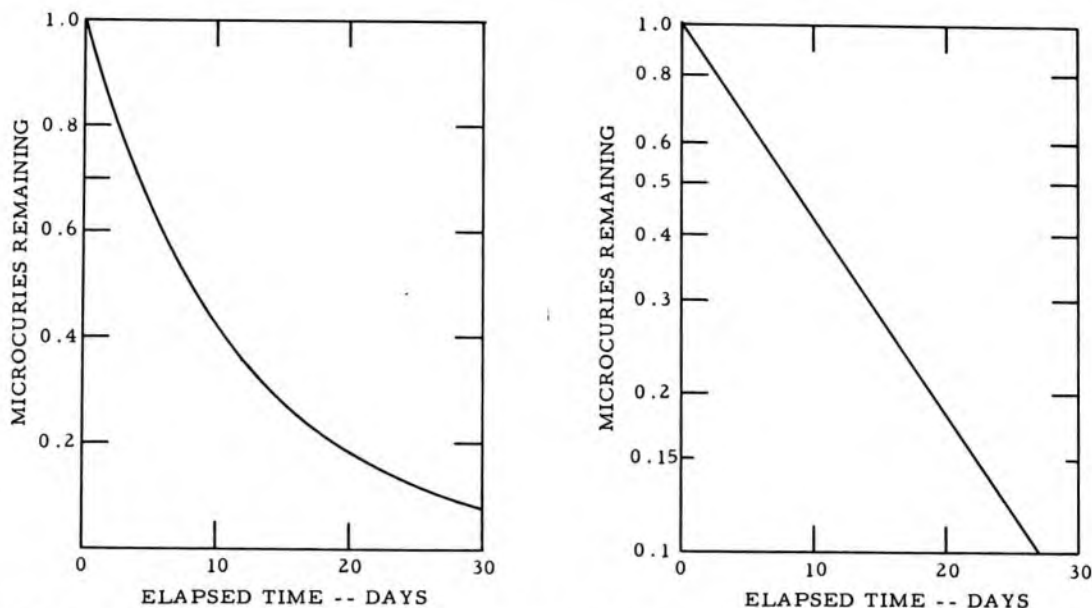


Fig. 4. Radioactive decay. Graphs show the disappearance of ^{131}I radioactivity with time, for an initial activity of 1 microcurie. Left. Linear scale graph paper. Right. Semilogarithmic graph paper; note that this is easier to read at lower activity levels.

Arithmetic of Radioactive Decay

As the atoms in a sample of a radionuclide disintegrate, the number remaining is correspondingly reduced. Hence, the radioactivity of a sample decreases continually with time. Careful measurements show the decrease follows an exponential law. Figure 4 shows this relationship plotted on both ordinary and semilogarithmic graph paper (for an initial activity of $1\ \mu\text{c}$ of ^{131}I).

Mathematically, one may express the relationship as:

$$F = e^{-\lambda t} \quad (9-14)$$

where F = the fraction of the initial radioactivity present after time t .

λ = the disintegration constant, characteristic of the radionuclide involved.

Note the similarity in form of Figure 4 and equation (9-14) to Figure 7 and equation (4-2) of Chapter 4, which refer to the transmission of narrow, monochromatic x-ray photon beams.

λ is defined implicitly by the above relationship, but it is rewarding to inquire further as to what it means physically. λ , the "decay constant," actually indicates the fraction of atoms of the radionuclide

TABLE 7. ¹³¹I REMAINING AFTER INDICATED NUMBER OF DAYS

Elapsed Days	Remaining Activity—nCi
0	1,000
8.1	500
16.2	250
24.3	125
32.4	62.5

present which decays per unit time. (More accurately, $\frac{dN}{dt} = -\lambda N$, so $\lambda = - \left(\frac{dN/dt}{N} \right)$.)

HALF-LIFE AND AVERAGE LIFE. Two related quantities are of greater importance practically: the “half-life” (T) and the “average life” (T_a) of the radionuclide. From Figure 4 and Table 7 it is evident that no matter when we measure the ¹³¹I activity, 8.1 days later half the material has disappeared. In general, one can define the physical half-life of a radionuclide as the time required for half of its atoms to disintegrate. This figure is unique for every radionuclide. (For ¹³¹I, T is evidently 8.1 days.)

How long do the radioactive atoms survive on the average? Evidently half of them will never survive a period T and, therefore, have lives from 0 to T in value; but others live longer, with a tiny fraction surviving a very long time. Calculation shows that, *on the average*, the atoms survive a time T_a, somewhat longer than T, given by the expression:

$$T_a = 1/\lambda = 1.443 T \tag{9-15}$$

The half-life T is so much simpler conceptually that one might ask, why bother at all with T_a? The answer is that T_a is essential to calculation of dosage, since it is used to estimate the total number of radionuclide disintegrations to which a tumor is subjected in a given decay interval.

BIOLOGIC AND EFFECTIVE HALF-LIFE. In biologic tracer studies one often desires to evaluate the rate at which material clears from an organ (for example, hormone secretion from the thyroid gland). One can often conveniently measure the decrease of radioactivity in this organ in vivo; this is related to the actual organ clearance time but also involves physical decay as well. In many organs there is a mass action law type of clearance so that a “biologic half-life” (T_b) exists for the organ. This can be defined as the time required for metabolism

of half of the *non-radioactive* form of the element. This is a purely physiologic quantity and is related to organ function only.

Actually, the measured radioactivity decreases with time as a result of *both* processes: excretion by the organ, at a rate characterized by T_b , and radioactive decay of all tracer material, at a rate characterized by T .

Evidently, the measured activity therefore decreases more rapidly than by either process alone. (This is similar to the more rapid clearance of money from a joint bank account when both people spend it.) The actual measured clearance has an effective half-life T_e , smaller than either of the others. By calculations one can show that $T_e = \frac{T T_b}{T + T_b}$, from which one easily obtains:

$$T_b = \frac{T_e T}{T - T_e} \quad (9-16)$$

It should be emphasized that exponential clearance from the organ is assumed. In some biologic situations transport is *not* passive, and this does not apply.

STATISTICAL NATURE OF RADIOACTIVE DECAY. So far we have discussed radioactive decay of relatively large numbers of atoms. In clinical tracer work one is limited by the need to minimize patient hazard to very low activities. Under these circumstances a new consideration becomes important: statistical fluctuations. These arise from the basic fact that the exact time an atom disintegrates is a matter of chance, and therefore the true remaining fraction of atoms fluctuates about the theoretical value of equation (9-14). This does not cause much trouble in measuring a large activity. Enormous fluctuations in apparent radioactivity, however, are obtained with low counting rates. Although we shall return to this question later, it is interesting to illustrate how statistics affect the actual measurement of radioactive decay. Figure 5 shows a theoretical curve for decay for ^{131}I (solid curve). (Shown are total counts for a fixed time interval.) The dotted lines indicate the likely range of variation in the actually measured counts at various times, for 20,000 total registered counts initially. More precisely, these lines encompass about 95 percent of the measured activity values, for a large number of measurements. Note the percent fluctuation increases rapidly at lower disintegration rates. We shall discuss statistical fluctuation further in Chapter 11.

Thus, equation (9-14) is an idealization which ignores statistical fluctuation. However, any statistical fluctuations occur about the values computed from equation (9-14).

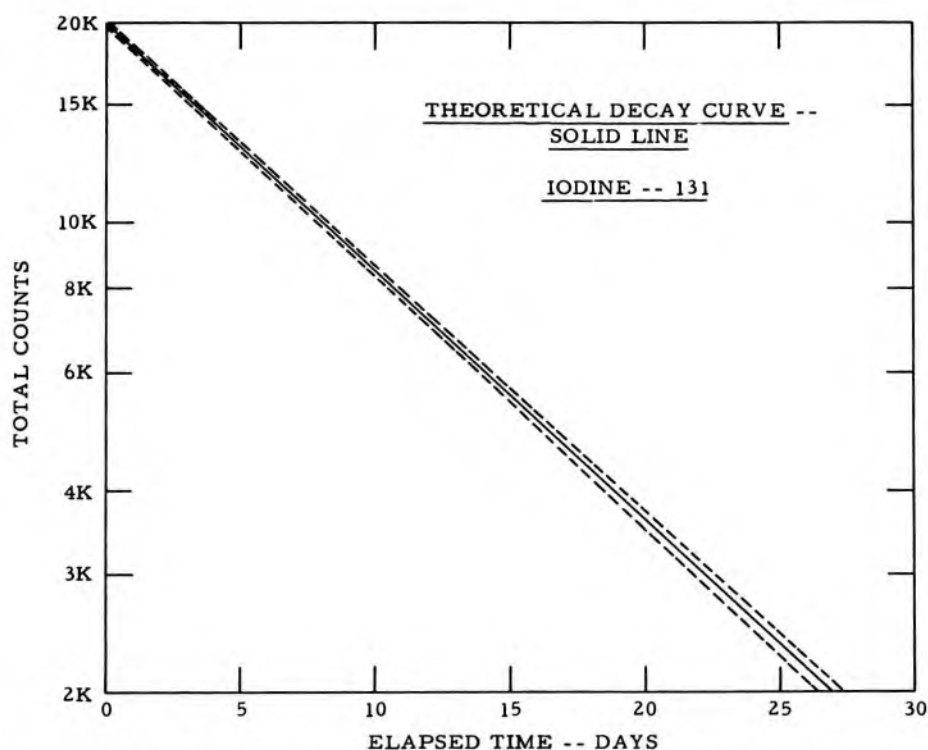


Fig. 5. Statistical fluctuation of measured activity. The solid line shows the expected reduction in total counts measured during a given time interval due to decay starting with 20,000 initially. This total should of course decrease with time as the material decays. However, as the total counts are reduced, increasingly great departures from the curve are noted in individual measurements. The dashed lines include 95% of the observed counts if one makes a large number of observations, according to probability theory. Note fluctuation increase as activity decreases.

MICROGRAMS REQUIRED FOR A MICROCURIE OF A RADIONUCLIDE. One microcurie is the activity of a microgram of radium element (Chap. 11). A microcurie of any other radionuclide, however, is always associated with a *different* mass. Table 8 shows the $\mu\text{g}/\mu\text{Ci}$ figures for several radionuclides, with their half-lives T and the product AT . Note that the $\mu\text{g}/\mu\text{Ci}$ ratio is proportioned to AT .

The $\mu\text{g}/\mu\text{Ci}$ is uniquely determined for *any* radionuclide by its (AT) value and is given by the expression:

$$m_0 = \left(\frac{A}{226}\right)\left(\frac{T}{1622}\right) \quad (9-17)$$

m_0 is the number of μg in a μCi of the radioisotope.

A is the atomic mass number (versus 226 for radium).

T is the half-life in *years* (versus 1622 for radium).

Table 8 also gives computed values for several useful radioisotopes.

TABLE 8. HALF-LIFE AND MASS PER MICROCURIE OF SEVERAL USEFUL RADIONUCLIDES¹⁰

A. Short Half-Life

Radionuclide	Half-Life T	AT*	Micrograms per Microcurie
³² P	14.3 days	1.254	3.43×10^{-6}
⁵¹ Cr	27.8	3.88	10.6
⁵⁹ Fe	45.1	7.28	19.8
⁵⁷ Co	270.	42.2	115.
⁷⁵ Se	127.	26.1	71.2
⁸⁵ Sr	64.	14.9	40.7×10^{-6}
⁹⁰ Y	2.67	0.658	1.795
^{99m} Tc	0.25	0.0678	0.185
¹²⁵ I	60.	20.5	56.
¹³¹ I	8.1	2.91	7.95
¹⁹² Ir	74.4	39.1	106.5×10^{-6}
¹⁹⁸ Au	2.70	1.465	4.
¹⁹⁷ Hg	2.71	1.46	3.99
²⁰³ Hg	47.	26.2	71.6
²²² Rn	3.83	2.33	6.36

B. Long Half-Life

³ H	12.3 years	36.9	0.000101
¹⁴ C	5,560.	77,800.	0.212
⁶⁰ Co	5.24	314.	0.000858
⁸⁵ Kr	10.3 years	876.	0.00239
⁹⁰ Sr	27.7	2,490.	0.0068
¹³⁷ Cs	26.6	3,640.	0.00994
²²⁶ Ra	1,622. years	366,500.	1.000
²³⁸ U	4.5×10^9 years	1.07×10^{12}	2.92×10^6

* A is atomic mass number, T is half-life in years.

Why should A and T be involved in this ratio? A involves the mass of each atom; certainly equal numbers of heavier atoms should weigh more than light ones, so this appears reasonable. Dependence on T is perhaps a bit more puzzling. The disintegration rate (R) is: $R = \lambda N$, or $N = R/\lambda$ (from 9-14). (N is the number of atoms present, and λ is the disintegration constant.) Hence N is proportional to both $1/\lambda$ and the radioactivity. Since $T = .693/\lambda$, the number of atoms needed to yield R is therefore proportional to the half-life T. The 226 and the 1,622 years in equation (9-17) are the A and T factors for ²²⁶Ra, for which m_0 is one μg per μCi by definition of the curie unit.

Millicurie-hour Concept

When radionuclides are used for therapy, tissue ionization results from energy absorbed from the emitted rays. Hence, tissue dose evidently depends on:

1. The number of disintegrations occurring during treatment.
2. The nature and energy spectrum of the rays from these disintegrations.
3. The geometry of the sources, i.e., where they are situated relative to the tissue treated.
4. Absorption of the emissions on the way to the tissue treated.

The number of disintegrations obviously depends on the activity in d/s or millicuries and on the length of time they occur in the patient. Since a millicurie is 37 million d/s, and an hour 3,600 seconds, evidently a millicurie in an organ for an hour exposes it to the rays from $(37,000,000) \times (3,600)$ disintegrations, or:

$$1 \text{ mCi-hr} = 133.2 \text{ billion disintegrations} \quad (9-18)$$

The mCi-hr is a convenient unit of exposure to radioactivity. If activity remains substantially constant during a given treatment time, total mCi-hrs are calculated readily by simple multiplication. Consider, for example, 90 mg of radium used to treat a patient for 60 hours. The radium activity is substantially constant during this treatment (a sealed radium source drops in activity less than one percent in 25 years because its half-life is 1,622 years), so the number of mCi-hrs = $90 \text{ mCi} \times 60 \text{ hrs} = 5,400 \text{ mCi-hrs}$. This simplified calculation is usually sufficiently accurate for clinical use whenever T is greater than the treatment time by a factor of 6 or 7.

Many useful radionuclides have half-lives more nearly comparable with typical treatment times. In such cases many fewer disintegrations occur than the above simple procedure would indicate because substantial activity is lost by decay during the treatment. A simple calculus integration results in a very convenient practical formula for handling this computation:

$$\text{the number of mCi-hrs} = (T_a) \times (\text{millicuries destroyed}) \quad (9-19)$$

T_a is the average life ($1.443T$) (see equation 9-5). The millicuries destroyed is exactly what the term implies: the loss in radioactivity during the treatment. In practice one can use equation (9-13) and exponential tables to calculate $(1-F)$, the fraction of the initial amount that decayed during the treatment. Generally, it

TABLE 9. TWO EXAMPLES OF CALCULATION OF MILLICURIE-HOURS

1. Basic formula: # mCi-hrs = (# mCis destroyed) \times (average life in hrs).
2. Example 1: 15 mCi of radon, left in place permanently.
 - (A) $T_a = T/.693 = [(3.83)(24)/.693]$ hrs = 133 hrs.
 - (B) All 15 mCi are destroyed since material is left permanently in patient.
 - (C) Hence, number of mCi-hours is: $(15)(133) = 2,000$ mCi-hrs.
3. Example 2: 20 mCi of ^{198}Au , left in place permanently.
 - (A) $T_a = T/.693 = (2.69)(24)/.693 = 93$ hrs.
 - (B) All 20 mCi are destroyed since material is left in patient.
 - (C) Hence, number of mCi-hrs is: $(20)(93) = 1,860$ mCi-hrs.

is more convenient to use semilogarithmic graphs or charts for this purpose. Table 9 presents two samples of such a calculation, for radon-222 and gold-198, two radium substitutes of relatively short half-lives.

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10

Basic Radioactivity Measurements

Measurements are made of emissions from nuclear processes for two general purposes. The first is to assure radiation safety, by evaluating dosage levels of both persons and work areas. The second is to study some process. This could be the nuclear reaction itself, or the physiology of a patient as manifested by the amount and distribution in the body of administered radioactive tracer material. Basic physics research and most fission and fusion applications involve sophisticated measurements of heavy particle and mixed beams of neutrons and gamma rays, which are beyond the scope of this book. We shall consider primarily clinical applications in this and the following two chapters, and radiation safety aspects later in Chapters 14 and 15.

This chapter is divided into four sections. The first discusses basic concepts of radioactivity measurement; the second and third, Geiger-Mueller (G-M) and scintillation detectors and their associated instruments; and the last, sources of errors and their control in radiation measurements.

RADIOACTIVITY MEASUREMENT CONCEPTS

Before considering the actual instruments used, it is important first to clarify some fundamental concepts of radiation measurement. These relate to the basic steps involved, the fraction of nuclear disintegrations actually registered, and the types of radiation detectors used.

Basic Steps

There are three steps in a measurement of radioactivity: detection, pulse processing, and display or "readout" of the result (Table 1).

Detection always starts with interaction of the particle with a gas, liquid, or solid to produce ionization. The ions may be collected

TABLE 1. STEPS IN RADIOACTIVITY MEASUREMENT

A. Detection	Ionization produced by passage of ray results in an electrical "pulse."
B. Pulse Processing	Three steps, generally: 1. Amplification 2. Selection 3. Counting.
C. Readout of Result	Many ways exist, including: 1. Scaler total displayed, with elapsed time. 2. Ratemeter direct answer in c/m. 3. Fast printout totals printed out, with corresponding time interval. 4. Scanning display of activity distribution in area of patient. Types include dot, color, photoscan, camera image, etc.

directly or employed to derive a larger electrical signal. The most useful detectors in medical work are ionization chambers, Geiger-Mueller tubes, and scintillation detectors. Some other detectors are also used for specific applications. Table 2 indicates several radioactivity detectors.

The resulting electric pulses may be processed in any of three ways. First, they may be amplified because of their initial low energy. Second, they may be sorted for pulse size in scintillation detection (see below). And finally, the number of pulses of each size is totaled to obtain the number of counts for a given time interval, and this result used to compute the detected count rate.

After counting, the result must be usefully displayed. Three types of instruments—scalers, ratemeters, and fast digital printout systems—are used.

A scaler is simply a totalizing instrument. It is used with an accurate timer to present the total number of pulses received during a given time interval. In one procedure, it is used to obtain total counts in a preset *time* interval (e.g., five minutes); this is "preset time" operation. Alternatively, one measures the time required to receive a preset number of *counts*; this is a "preset count" operation. The former procedure assures convenient times of measurement; the latter, desired statistical accuracy (see below). The average count rate is obtained in either method by dividing the total counts by the elapsed time. Modern units display the totals conveniently and accurately with minimum difficulty, usually on glow tube regis-

TABLE 2. RADIOACTIVITY DETECTORS*

A. Tissue Dosage Measurement (roentgens and rads)	
1. Ionization chambers	most accurate; major clinical instruments
2. Fluorods	in-vivo radium dosimetry
3. CdS crystals	in-vivo radium dosimetry (checking radium source placement)
4. Fricke dosimeter $\text{Fe}^{++} \rightarrow \text{Fe}^{+++}$.	high energy dosimetry (1 MeV and over)
5. X-ray film	high energy beam distributions
6. LiF dosimeters	tiny size in-vivo and research dosimeters
B. Count Rate Measurement (c/m)	
1. Geiger-Mueller (G-M) tubes† . . .	laboratory monitoring; moderate to high energy beta ray counting
2. Scintillation counting†	most accurate gamma and low energy beta counting
3. p-n junction diodes	Li-drift Si—particle counting. Li-drift Si and Ge—gamma counting, for high resolution spectrometry.

* See Tables 1 and 2 in Chapter 6.

† These are discussed at length in the text.

ters. In older units mechanical and slower electrical display systems often responded inaccurately to rapid sequences of pulses. To prevent missed counts, “scaling circuits” were employed to “scale down” the pulse rate; in this procedure only a fraction of the incoming pulses was conveyed to the registering circuit; the number was reduced by a “scaling factor,” usually 2, 4, 8, 16, etc. or 10, 100 etc. This is how the term “scaler” originated. Scalers are basic machines in radioactivity measurement work.

Sometimes one desires to know the count rate variation of rapidly changing radioactivity. (Such a change could be from physiologic processes or the use of very short half-life radionuclides.) Ratemeters provide a direct readout of count rate on a meter or recorder and are often very convenient. The circuit used to compute the count rate from the pulse sequence, however, introduces its own errors, as does the meter or recorder. The ratemeter, therefore, generally provides the advantage of convenient indication at the cost of reduced accuracy.

Digital printout units, essentially rapidly recording scalers, are now available. They print total counts and the corresponding times at desired intervals. Amazingly rapid mechanical printing is now possible, originally developed for computers. In most such units the user must compute the average counting rate, but computation and even curve plotting options are also available. Generally, such systems can provide greater accuracy and convenience, for a price.

To this point we have considered measurements at a single location. Often one wishes to know the activity *distribution* over an entire area of the body—thyroid, lungs, brain, liver, spleen, kidneys, etc. Various instruments and techniques are employed to perform such “scanning.” The results may be displayed as dot sequences on paper (“dot scans”), photographic images (“photoscans”), and the newer Polaroid film images of cathode ray oscilloscope screens (“camera scans”). This very important topic is discussed further in Chapter 11.

Counting versus Disintegration Rate

The purpose of a measurement is, of course, to obtain the activity, or *disintegration rate*. But *counting rate* generally differs greatly from this: a one microcurie source disintegrates 37,000 times per second, but normally far fewer counts per second are registered. Four variables contribute to this difference; they are geometry, attenuation, contributory scatter, and detector response (Table 3).

TABLE 3. FACTORS AFFECTING COUNTING RATE FOR A GIVEN DISINTEGRATION RATE (See Figure 1.)

A. Geometry	Many rays miss the detector altogether. Distance and detector and source size and shape are all involved.
B. Attenuation	Rays are attenuated on the way to the detector. This can be in the sample or organ itself or in intervening material such as other tissue or a container, air, and the detector cover (“window”).
C. Contributory scatter	Rays initially directed <i>away</i> from the detector can be deflected <i>towards</i> it. This can be by any mounting container or other surrounding material.
D. Detector response	Many rays entering the detector do not produce a detected pulse. The fraction of entering rays actually counted is called the “detector efficiency.”

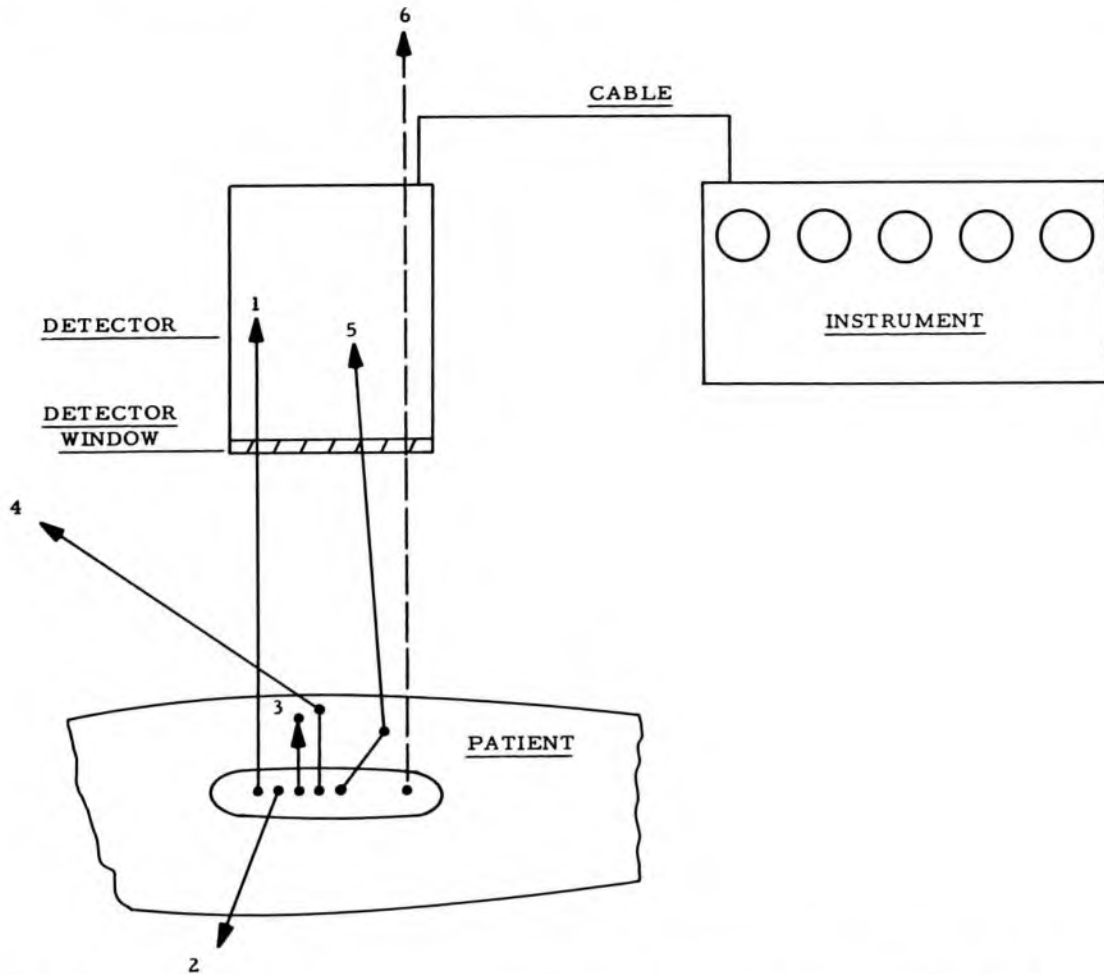


Fig. 1. Counting vs. disintegration rate—in-vivo measurement. Only a fraction of the emitted rays are registered. Six possible events are illustrated. 1. Ray enters directly from the source and is counted. 2. Ray is in wrong direction; misses detector altogether. 3. Ray is absorbed on way to detector (absorption). 4. Ray is scattered away from detector (deflection attenuation). 5. Ray initially directed away from detector is scattered into it (contributory scatter). 6. Ray strikes detector but produces no response (dotted line), due to inefficiency of detector.

Figure 1 illustrates the situation for gamma photon detection in vivo. Photon 1 is directed initially towards the tube and results in a registered count. Photon 2 misses the detector altogether and has no chance of being counted. For the situation shown, very few rays are directed towards the detector, so most disintegrations cannot possibly be counted. The detector size and distance from the source obviously affect geometry. Photon 3 is absorbed by intervening tissue, and photon 4 is deflectionally attenuated. Attenuation, of course, could happen anywhere along the way: in the concentrating organ, intervening tissue, air, and the detector window. To some extent contributory scatter partly compensates for attenuation, as indicated by photon 5, which is counted even though it was initially

directed to miss the detector. Finally, note that photon 6 is not counted even though it reaches the sensitive detector volume. In general, only a fraction of the rays entering the detector is registered; this fraction is called the "detector efficiency." For gamma rays, it is generally well below 2 percent for most G-M tubes but between 30 and more than 90 percent for scintillation crystals. Specifically designed G-M tubes, however, have very good *beta ray* efficiencies.

Detectors

A wide range of detectors is available for various measurements, as indicated in Table 2. For most x- and γ -ray therapy, ionization chambers provide the most accurate *dosage* data. However, all the other detectors are useful for specific problems.

For medical radioactivity measurements, G-M and scintillation counters are employed to detect individual beta and gamma particles. G-M tubes are very useful for beta-ray counting, but scintillation crystals are most often used for gamma-ray work.

GEIGER-MUELLER TUBES

Until relatively recent improvements in scintillation units, Geiger-Mueller, or G-M tubes, were the most widely employed radia-

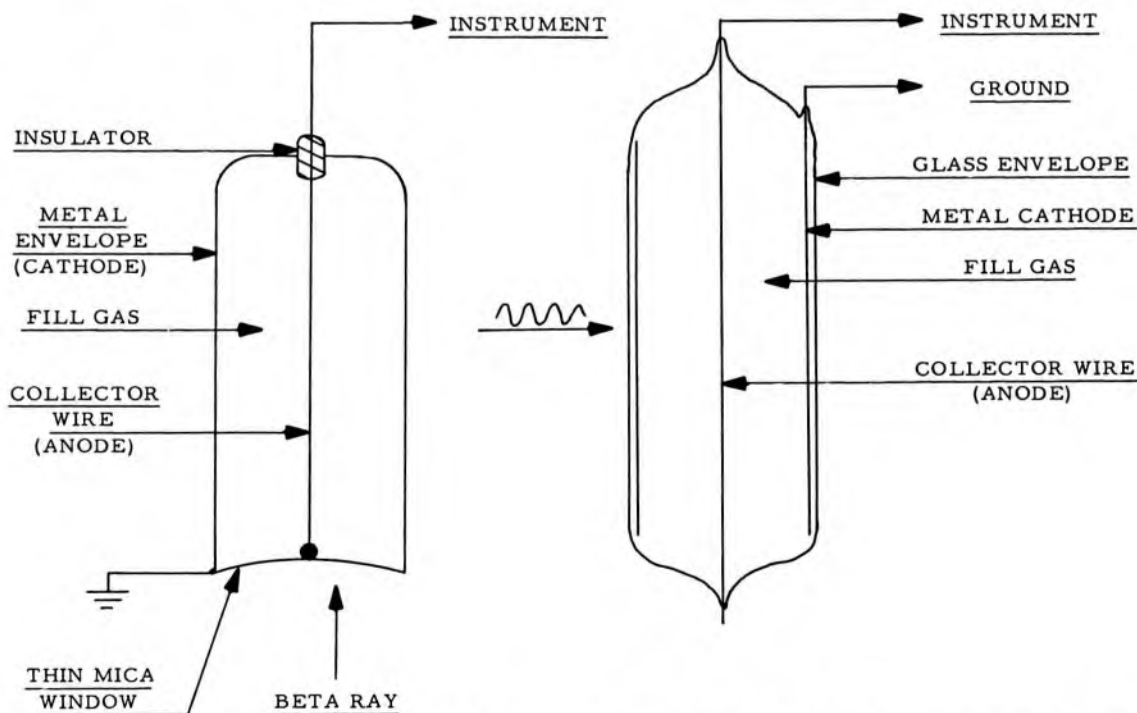


Fig. 2. Two common types of Geiger-Mueller (G-M) tubes. Left. End window tube. Right. Side window tube.

tion detectors used in medicine.^{1, 2} They are still very useful, particularly for radiation safety survey and beta-ray measurements. The more sensitive scintillation crystal detectors are almost universally preferred for gamma-ray tracer studies, however.

A G-M tube consists basically of a sealed container of gas with two electrodes to collect gas ions. The exact design depends upon the type of particles to be detected and other aspects of the desired use, but all G-M tubes have a thin positive electrode, usually a very pure tungsten wire only a few mils in diameter. (Figure 2 shows two common types.) It will be recalled that an ionization chamber is also a gas-filled container with two electrodes. But there is great difference in performance—a G-M tube can be millions of times more sensitive to radiation than an ionization chamber of the same size. This surprising fact stems basically from the much greater electric field strengths employed in G-M tubes than in ionization chambers.

Basic Operation

The best way to approach G-M tube operation is to consider a basic ionization chamber as its collection voltage is gradually increased from zero to high values. We consider below what happens as this is done, until G-M tube discharge occurs. We then discuss “quenching,” or terminating such discharge.

EFFECT OF VARYING COLLECTION VOLTAGE. Consider Figure 3 showing a gas tube C subjected to a source of constant intensity radiation. This produces primary ions at a constant rate, which one would

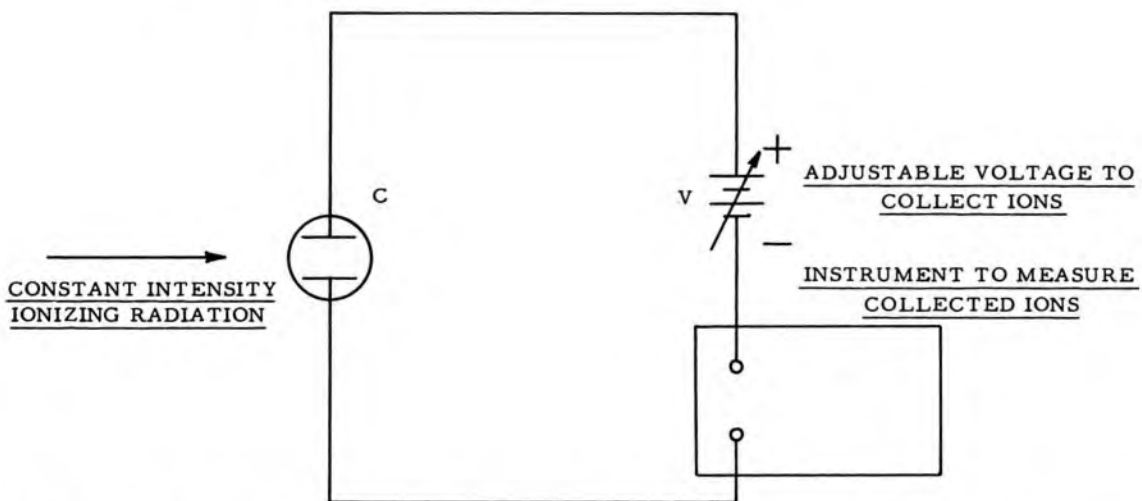


Fig. 3. Gas ionization system with adjustable charge collection voltage. A source of constant intensity radiation is employed, assuring a constant rate of production of gas ions before V is applied. The collected charge is now measured as the voltage is increased from zero. Figure 4 is a graph of the result.

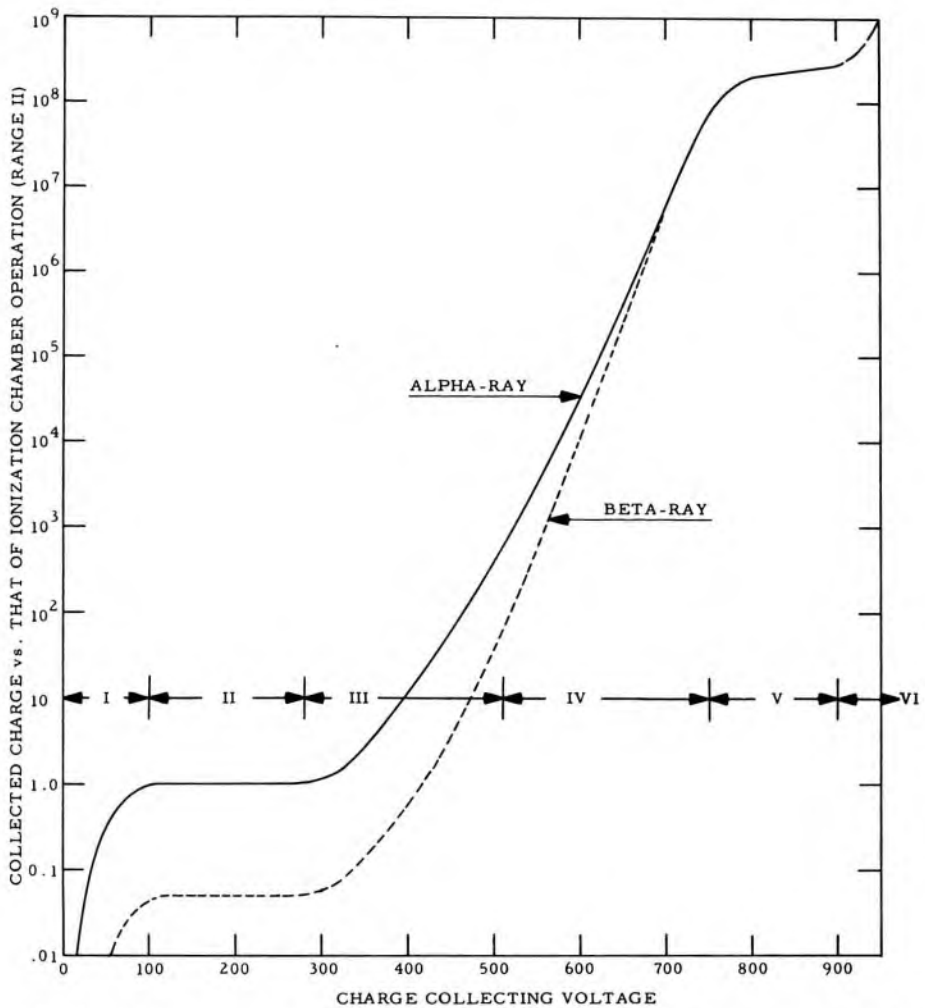


Fig. 4. Collected charge vs. that for ionization chamber operation. As voltage is varied in Figure 3, six ranges are observed: I. *Low voltage*. Ion recombination region. II. *Saturation voltage*. Substantially complete collection of radiation-produced ions. Ionization chamber region. III. *Higher voltage*. Ion multiplication. Proportional region. IV. *Even higher voltage*. Greater ion multiplication, to extent pulse differentiation is less easily accomplished. Limited proportional region. V. *Plateau voltage*. So much ion multiplication that collected charge is independent of original ionization. G-M region. VI. *Very high voltage*. Gas and/or other insulation becomes unreliable. Spurious pulses, sparks, etc. result. Continuous discharge region. Note the separation of the alpha- and beta-ray curves from I through III permits ready distinction between alpha and beta rays; in IV it is harder to accomplish, and impossible in the G-M region.

normally expect to yield a constant collected current reasonably independent of the collecting voltage. Actually, this happens for only a certain limited voltage range. Figure 4 shows the actual variation of collected current versus collecting voltage V , for a particular G-M tube.

Six ranges of voltage are shown in the figure, each with its own characteristic tube behavior. I and II have been previously con-

sidered in connection with ionization chambers; III and IV are the “proportional region”; V is the G-M region; and VI is the self-discharge region. Let us consider these voltage ranges in more detail. Figure 5 shows the locations in the tube where collectable ions are produced for three different applied voltages. In each case, we assume a gamma photon is absorbed by the cathode surface at E, liberating a photoelectron or recoil electron. This particle ionizes gas in its path, which is shown as dashed line EA in all three cases. Now consider what happens in the tube as the collecting voltage is gradually increased.

IONIZATION CHAMBER OPERATION (REGIONS I AND II). At low voltages, collected ions all come from interaction of the original photoelectron or recoil electron with the gas. In Figure 5(Left), ions from a particular photoelectron all originate along EA and are collected as shown by the small arrows. It is clear the dashed lines completely enclose the volume of gas in which ions flow.

As previously shown in Chapter 6, substantial ion recombination may occur in the very low voltage region I because the electric field is small. In region II almost all ions produced are collected; this is the familiar voltage saturation region. With voltage saturation, the

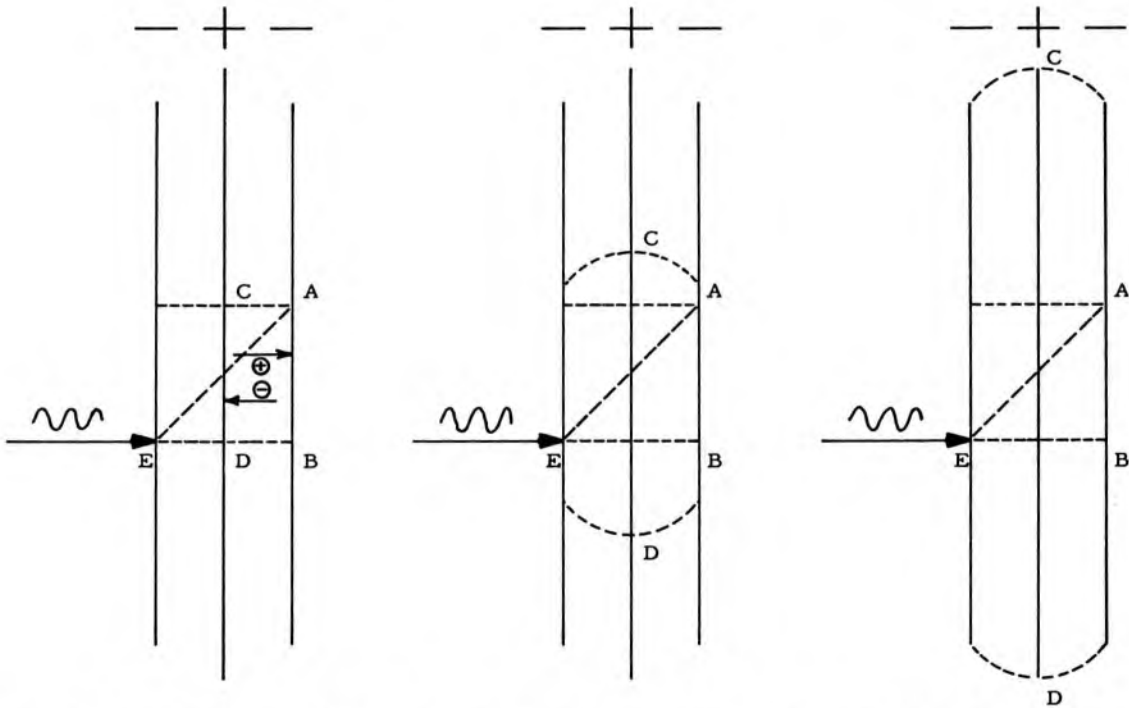


Fig. 5. Location of discharge in gas ionization tube vs. original electron track (diagonal line EA). Left. *Low voltage, ionization chamber operation.* Collected changes all come from essentially the volume between A and B. Center. *Proportional operation.* Photoelectrons released by center wire augment charge volume; discharge spreads to C-D along wire but does not fill entire volume. Right. *G-M operation.* Entire gas volume of tube which is in strong electric field glows.

current accurately indicates the rate of ionization from the incident radiation, since substantially all produced ions are collected and no other source of ions exists in the tube.

PROPORTIONAL COUNTER OPERATION. When the collection voltage is further increased a great increase in tube current is soon noted (region III). The reason for this is evident from study of Figure 5(Center). Electrons produced initially along EA acquire considerable energy as they move towards the positive collector wire, enough so they in turn ionize gas atoms they strike. Some of the electrons released from the latter atoms may in turn ionize *other* atoms. This supplementary ionization occurs with increasing frequency as the voltage is increased. As a result, the collected current is increased by ion multiplication; this increase can be of the order of 100 to 10,000 times; the extent is critically dependent on the collecting voltage. The collected charge is *still* dependent on the original ionization along line EA, as shown by the roughly parallel solid and dotted curves in Figure 4, for an alpha and beta ray, respectively. This is the *proportional* region, so called because the produced total charge is still roughly proportional to the initial ionization.

Note in Figure 5(Center) that proportional counter operation involves a moderate expansion of the ion collection location (CD) in the tube.

G-M COUNTER OPERATION. At higher voltages (range IV of Fig. 4), tube currents from the alpha and beta ray become more and more nearly identical; also enormous current amplifications are noted— 10^8 (one hundred million times).

What is responsible for this change in tube behavior as voltage is increased from range III through IV? A new effect starts to dominate the picture as the voltage increases: energetic light photons are released in large numbers by ionization of gas molecules. The subsequent events are rather complicated and dependent on both tube design and details of the electrical circuit used. However, these light photons (particularly ultraviolet ones) release photoelectrons from the collector [Fig. 5(Right)]. If the photons are not absorbed on the way, they release electrons from the cathode as well. This effect hence introduces a new supply of electrons; gas conduction now takes place at areas remote from the original ionization site because photons may illuminate the full tube length (CD).

With a great enough collecting voltage, the entire gas volume between electrodes becomes conductive in a self-accelerating process, rather appropriately called an "avalanche." The tube current reaches a constant maximum value regardless of the original ionization. For example, the same final tube current results from a single meson

which directly produces only four ion pairs, and from an alpha ray which directly produces thousands of ions.

This region is shown as V in Figure 4 and is called the G-M region. Note in the figure that at 850 V the tube produces a charge pulse roughly 200,000,000 times greater than is produced at 200 V, when using the tube as a simple ionization chamber. The G-M charge amplification is seen to be enormous (more than 1,000 times that of even proportional operation).

Between III and V is a transitional region of "limited proportionality." Here the U-V photon effects gradually increase with voltage, being very low in III and dominant in V. Beyond V, the voltage is large enough for unstable operation or continuous discharge to occur. *Operation of a G-M tube beyond the G-M region greatly reduces its life and can potentially destroy it outright.*

To summarize, ionization chambers and proportional counters give a direct measure of the initially absorbed energy. They can therefore be used to differentiate between an alpha and beta particle because of the higher LET in the counter of the former. The resulting total collected charge, however, is extremely small. Ionization chambers are entirely too insensitive for tracer work. Even proportional counters require very high gain amplifiers and extremely stable voltage supplies, normally limiting their use to research laboratories.

The G-M counter's great advantage is that a relatively large charge pulse is produced by *any* ionizing event in the tube. This makes the required circuitry following the tube quite simple and inexpensive.

G-M COUNTERS—QUENCHING. In ionization chambers and proportional counters the charge pulse disappears almost at once, as soon as the detected ionizing particle gives up its energy. This is because current flows only during production of ions by the detected particle; when this is over, so is the tube current. Also, currents are tiny, so the ions do not significantly affect the electric field.

G-M counters are different. Once triggered by the original ray, they tend to continue to discharge until turned off by another mechanism. Evidently, without quenching of the discharge the tube can respond to only a single ray; after this it continues to discharge and is therefore insensitive to subsequent rays.

Two basic quenching methods have been employed: external (electrical) and internal (chemical). Historically, electrical circuits were used first, usually with tubes containing argon or nitrogen and hydrogen gas. These circuits automatically reduce the tube voltage shortly after the avalanche starts, halting ion production almost instantly; the remaining ions are then gathered up by the reduced

collecting voltage. Full voltage is restored automatically after quenching and the tube readied for the next pulse.

Chemical quenching is most generally employed today in G-M tubes because the setup is more dependable for routine work. The filling gas is usually a mixture of argon and ethyl alcohol, ether, or amyl acetate. The organic molecules absorb ultraviolet light very effectively, so the light produced by electron collision is absorbed on the way to the tube cathode. However, a considerable number of ultraviolet photons strike the collector wire because they are produced in the strong electric field region near this wire. Full ionization occurs near the wire, with a cloud of positive ions migrating outward relatively slowly towards the cathode. The presence of the quenching molecules near the collector wire combines with the accumulation of positive ions nearby to extinguish the discharge after about 50 to 150 μsec , and the positive ions are then swept away, preparing the tube for a new discharge.

Unfortunately, dissociation of organic quenching compounds occurs when they are struck by U-V light and electrons. This is irreversible, so the material is consumed a bit with each count. The life of these tubes is hence limited to between 1,000 and 10,000 million counts, depending on the tube design and use. One should obviously not waste G-M tube counts by leaving the high voltage supply turned on unnecessarily. Some tubes use chlorine or other halogen gases for quenching; these recombine spontaneously, so such tubes have longer lives. They are, however, generally somewhat less sensitive.

Special Considerations in Use

Having discussed the basic nature of G-M tubes and the production of electrical pulses, we now consider several vital and unique aspects of their use: the voltage plateau, resolving time, and instrument requirements.

VOLTAGE PLATEAU. We showed in Figure 4 how the pulse size varies with tube voltage. The proper G-M operating region would be about 850 V for the tube shown. It is, however, quite difficult to measure a curve like that shown in Figure 4 and much easier to determine the "characteristic curve" shown in Figure 6. This is a graph of measured *counting rate* versus tube voltage, for a constant activity source. As the voltage rises from zero, no counts are detected until a fairly high voltage (called the "starting potential") is reached. The count rate then rises rapidly and levels off beyond a "threshold." A relatively level portion called the "plateau" includes the proper operating point of the tube. When the voltage is excessive, operation becomes unstable

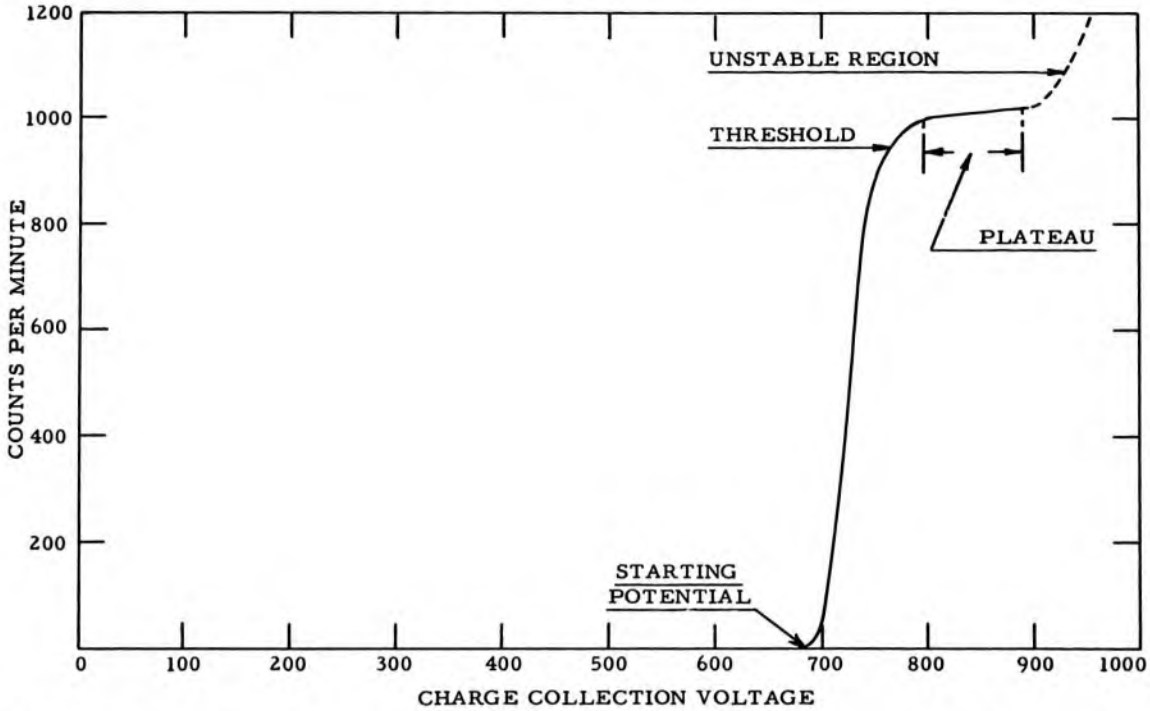


Fig. 6. G-M tube characteristic curve. This relates the measured counting rate in c/m to the collecting voltage, for a constant intensity irradiation of the tube. Some definitions: Starting potential: required potential before significant c/m occurs. Threshold: start of plateau. Plateau: relatively flat portion of curve, for which the count rate is relatively independent of the tube voltage. Continuous discharge region: above the plateau unstable operation occurs, and tube is readily damaged. The best operating point is about one-third up on the plateau. This assures plateau operation even with substantial line-supply change and is well below the continuous discharge region.

in the “continuous discharge region,” where erratic quenching and other problems arise. The operating voltage should be selected near the low voltage end of the plateau for maximum tube life.

What causes the peculiar shape of the characteristic curve? The unresponsive region below the starting potential corresponds to production of very tiny electrical pulses below the G-M region. Recall that the vertical scale of Figure 4 is logarithmic; it is evident pulses are very small even at 700 volts, and the subsequent instrument simply cannot respond to them at G-M operation settings.

With voltages beyond the threshold value, all pulses are full size and therefore readily counted. Along the plateau, there is a marked independence of pulse size and hence counting rate on voltage. As a result, the measured count rate with a G-M tube is relatively independent of small changes in G-M tube voltage.

The plateau steepness is an index of G-M tube quality. Generally, there should be only a few percent change in counting rate with a 100 volt change in collecting voltage along the plateau. This percent

change in count rate per 100 volts is usually referred to as the "plateau slope." The slope increases with tube age and use; at least part of the cause is loss of quenching gas.

RESOLVING TIME—NATURE. There are unfortunately several disadvantages to G-M tubes. One of the most serious is their relatively great resolving time. This is defined as the required separation in time of two consecutive pulses for the second one to be counted. When pulses arrive too frequently for all to be detected, we refer to the resulting loss of some counts as "coincidence loss." All radiation detectors and their associated equipment have some such losses because of their finite counting speed. Of all detectors G-M tubes have the highest coincidence losses because their resolving time ranges from a minimum of about 50 through 300 or 400 microseconds. (The same tube used as a proportional counter is much faster.)

The resolving time consists of two parts: dead time and recovery time. Dead time is the time during which the tube discharges, until quenching ends active discharge. During this time the tube is unresponsive, or "dead," because it is already discharging. The ending of the discharge does not mean the tube is ready for action, however. The positive ions are heavy, move relatively slowly, and must be completely collected before the strong electric field is fully restored to ready the tube for the next pulse. The time to sweep enough of the ion "debris" away to permit response to the next pulse is the "recovery time."

What determines the resolving time? Dead time is controlled mainly by the quenching speed. Recovery time is obviously dependent on both ion speed and the distance ions must travel—i.e., the gas molecular weight and the tube size. The required pulse size for registration by the instrument is also involved.

RESOLVING TIME CORRECTION. A 300 microsecond resolving time requires a correction of 0.5 percent per 1,000 c/m actually measured. For example, an observed 3,000 c/m is truly 1.5 percent greater, or 3,045 c/m; 10,000 c/m measured is 5 percent greater, or 10,500 c/m. More generally the *correct* counting rate R is given by:

$$R = \frac{r}{1 - rT} \quad (10-1)$$

where: r is the observed count rate (c/m).

T is the resolving time (in minutes).

Another important aspect of resolving time is the failure of a G-M detector in very strong radiation fields. For example, a G-M

detector near a strong radium source may actually *decrease* in reading as one approaches the source because the G-M tube is overwhelmed by the excessive intensity (i.e., the tube remains discharged almost continuously, reducing the counts actually registered).

INSTRUMENT REQUIREMENTS. The pulses from G-M tubes are substantial in magnitude and can be connected directly to most instruments using a standard coaxial cable. An exception is some externally quenched tubes using a preamplifier stage near the G-M tube as a quenching tube (i.e., in the Neher-Pickering circuit). Normally the cathode is at ground potential, and the collector wire is connected to the high voltage through a large resistor (of the order of 1 to 2 megohms) (Fig. 7). During discharge the collector wire potential drops momentarily from roughly 1,000 V (the full supply voltage) to a few hundred volts or less; the resulting negative pulse of several hundred volts is coupled to an amplifier in the main instrument, which then processes it appropriately.

The instrument thus need supply only a reasonably stable and adjustable high voltage to the tube via a single coaxial cable. This voltage normally is readily adjustable from zero up to 1,500 V, to permit checking the plateau regularly. A simple relatively inexpensive voltage stabilizer circuit is adequate since the plateau slope is usually only a few percent per 100 V.

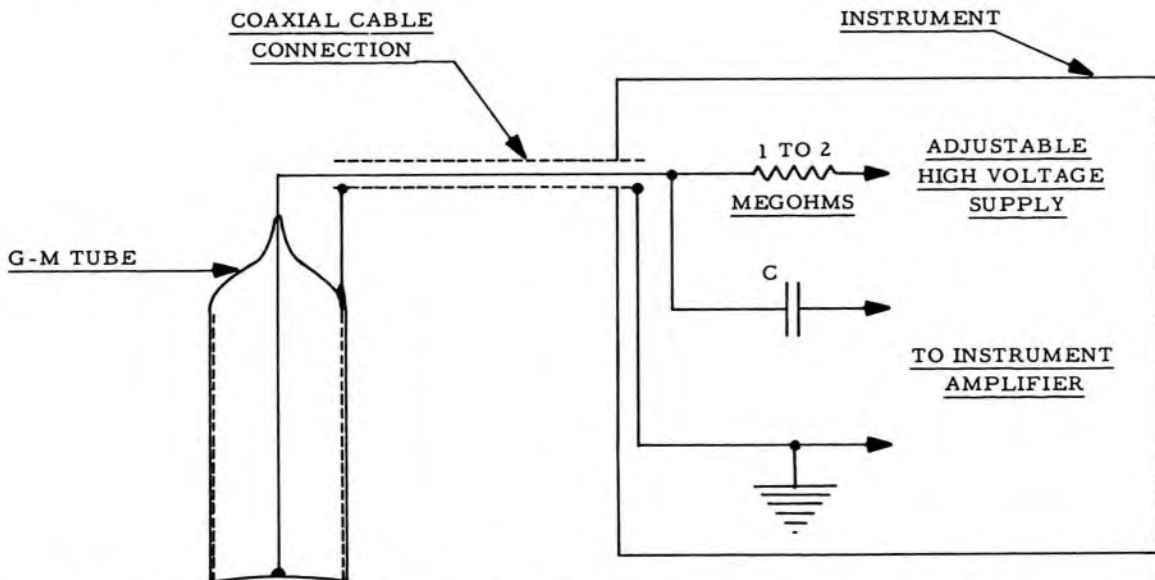


Fig. 7. Electrical connection of G-M tube to its associated instrument. A single coaxial cable conveys high voltage to the G-M tube, as well as the G-M pulses to the instrument amplifier. The discharge reduces the voltage across the tube from full to a relatively low value because of the IR drop across the 2 megohm resistor. The resulting pulse is communicated to the amplifier through a capacitor C, which also blocks the high voltage from the amplifier's first tube.

Design and Uses

G-M tubes have been used for counting both beta and gamma rays. We shall briefly discuss these uses below and describe the essentials of several G-M tube designs.

BETA-RAY COUNTING. Many biologically important elements do not have available gamma-emitting isotopic forms. These include hydrogen, carbon, oxygen, phosphorus, sulphur, and chlorine. Some important organic compounds preserve their biochemical behavior when a hydrogen atom is replaced by a gamma-emitting ^{131}I atom; most compounds, however, do not. Consequently, many research and some clinical studies employ beta-emitting radionuclides, of which ^{32}P , ^{14}C , and ^3H are perhaps most representative. Three different detectors may be used to measure beta-emitting tracer materials: end window G-M counters (^{32}P), gas flow G-M counters (^{14}C), and liquid scintillation counters (^3H and ^{14}C). We discuss the G-M counters here and liquid scintillation counters in the next section.

END WINDOW G-M COUNTERS. Figure 8 shows an end window tube beta counting arrangement. (Generally, a surrounding lead container is employed to minimize background counts.) The tray, or "planchet," is mounted on a thin recessed plastic plate, which can be reproducibly positioned at several desired distances from the tube window. The sample must be dried before counting to minimize absorption. (Recall that even ^{32}P rays are 50 percent absorbed by

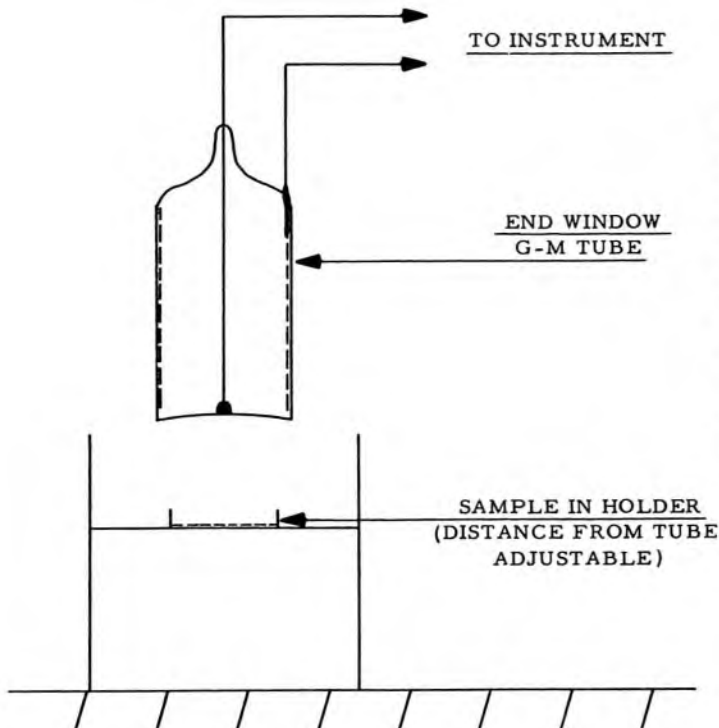


Fig. 8. End window tube beta counting arrangement. This simple method is usually limited in application to measurement of fairly energetic beta emitters, for which beta absorption and scatter errors can be controlled with reasonable effort. Note: Generally a lead shield surrounds the G-M tube to reduce background (not shown).

2.5 mm of water.) Absorption effects must be considered even with dried samples. Scatter of beta rays can also introduce problems because beta rays initially directed *away* from the tube can be scattered back into the tube. The amount of such scatter depends on the type of planchet used (metal versus plastic and its thickness) and the sample area, among other things. As a general rule, in beta counting of dried samples one must standardize the total mass and area of sample, the geometry, and type of planchet. A standard reference activity is needed for best accuracy. By contrast, gamma well scintillation counting is relatively simple to carry out accurately (see the next section).

GAS FLOW G-M COUNTERS. End window counters are not very satisfactory for measuring radioactivity of low energy emitters like ^{14}C and ^{35}S ; they are useless for ^3H beta rays. ^{14}C and ^{35}S beta rays are successfully counted in gas flow G-M counters (Fig. 9). In such units the problem of detector window absorption is solved by putting the dried sample *inside* the tube. (The geometry is thereby also greatly improved. These are sometimes called 2π solid angle detectors. Special versions are made for 4π solid angle detection to detect practically *all* the beta rays. The latter are useful for absolute measurements of disintegration rates.) Dry G-M counter gas from a tank is bubbled through the measuring (M) and storage (F) chambers at the rate of about 1 to 6 bubbles per second, under essentially atmospheric pressure. The gas flushes out air and moisture, and reproducible and accurate readings are obtained if one employs reasonable experimental care. Note the second and third samples are prepared during measurement of the first. This expedites measurements, but sample handling and flushing still limit the number of samples one can count in a given period.

Contamination of the counter must be carefully avoided, as well as dirt or grease on the charge collecting wire and insulation. It must be kept in mind that ^{14}C and ^{35}S beta rays are easily absorbed; great care is required in sample preparation. (Fifty percent of ^{14}C beta rays

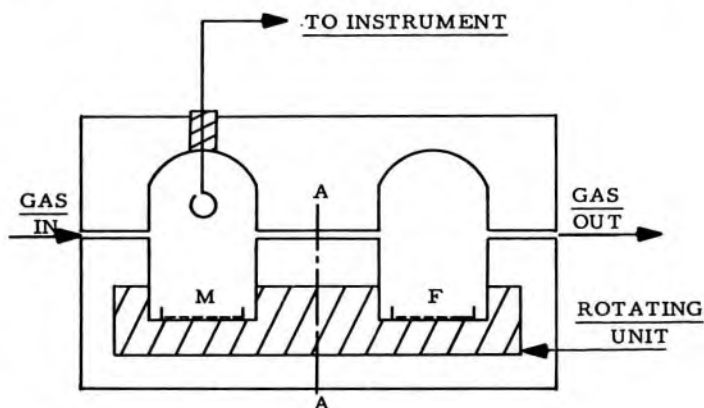


Fig. 9. Gas flow counter: basic principle. The rotatable unit (shaded) holds both the sample being measured (M) and two others being flushed (F, only one shown). Assembly rotates about AA to insert new sample into counter.

are absorbed by 0.043 mm of water or soft tissue.) Self absorption is less serious in liquid scintillation counting because the material is dispersed in the solution containing the detecting scintillating compound. This fact, in addition to its greater possible automation and speed, has made modern liquid scintillation counting the preferred method for virtually all beta sample counting. (See the next section.)

PORTABLE G-M BETA-GAMMA TUBE. End window G-M tubes are very useful for portable radiation safety survey instruments (Fig. 10). The thin end window permits beta rays (alpha also, if extremely thin windows are used) to enter the sensitive volume, where they are efficiently detected. Gamma rays can be detected from photon interactions with the glass envelope and cathode, which yield Compton recoils and photoelectrons to initiate the discharge. Normally gamma rays are counted rather inefficiently (about 1 percent or less) by such tubes. They are sufficiently sensitive, however, to be very useful for general laboratory survey use.

EXTRA-SENSITIVE GAMMA-RAY G-M DETECTORS. At present almost all gamma-ray measurements are performed using scintillation detectors. Modern large scintillation crystals routinely respond to gamma rays with efficiencies approaching 100 percent for typical radionuclide photons and well above 30 percent for even very high energy gamma rays. Consequently, clinical scintillation counting measurements

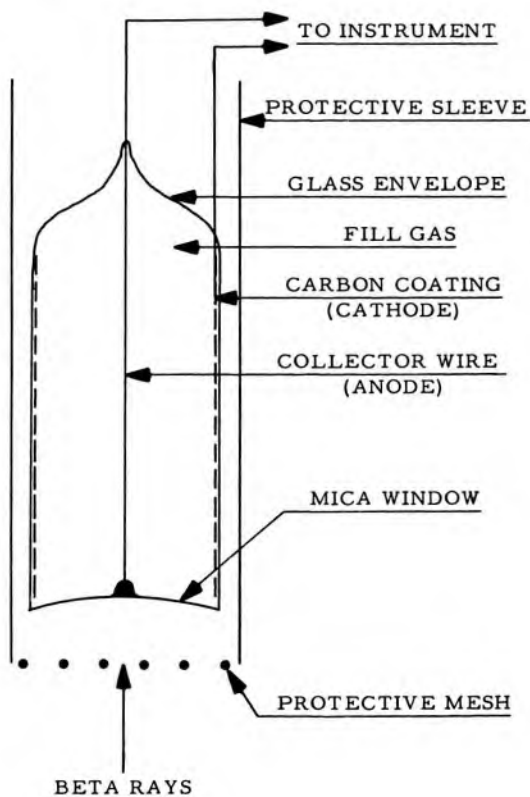


Fig. 10. End window tube as used in portable radiation protection survey meter.

routinely require 5 to 20 times less administered radioactivity than G-M counting measurements, a great benefit in reducing patient hazard.

For completeness we mention some special sensitive gamma-ray G-M detectors developed about 1950. These have efficiencies as great as 8 to 10 percent for ^{131}I gamma-ray detection. They employ high Z cathode materials with special internal construction to increase the total number of photoelectrons ejected into the gas volume by high energy gamma rays. One useful design uses walls of stainless steel, another, bismuth deposited on the walls.

SUMMARY. When applicable, G-M tubes have some advantages over scintillation detectors, including lower cost of both the detectors and instruments, simplicity and reliability of the system, and convenient intermediate and high energy beta counting of small numbers of samples.

They have corresponding disadvantages, however. Their sensitivity to gamma photons is quite low (10 percent or lower efficiency), and they cannot discriminate among α -, β -, and γ -rays or among rays of a given type with different energies. Furthermore, they have substantial resolving times and limited tube life. Scintillation systems, on the other hand, are more sensitive, readily discriminate among particle energies, and have both low resolving time and practically unlimited life.

SCINTILLATION DETECTION

We have so far described only gaseous ionization detectors in which the derived signal results from ion collection. Scintillation detection is even more important to medical tracer work.⁴ As early as 1899, H. Becquerel found that ionizing radiation produced luminescence in certain materials like zinc sulphide, barium platinocyanide, and diamond. By 1908 it was established that this light is produced by individual α -, β -, γ -, or x-ray particles. Thus, by counting light flashes or "scintillations," one can evaluate radioactivity just as with gaseous detectors. In addition, the light intensity is roughly proportional to the energy originally absorbed by the scintillator. This permits distinguishing among α -, β -, and γ -rays and among x- and γ -rays of different energies.

Signal Production

As with all radiation effects, the detected particle is first absorbed by the scintillant liquid or crystal involved. These units differ

basically, however, from gaseous detectors in that scintillations follow excitation on a crystal or molecular level, rather than ionization. When the crystal or molecule returns to the unexcited state, a definite amount of fluorescent energy is released. It can be as heat or some other unproductive form or as single or multiple light flashes. If the material is transparent enough, the light escapes and can be registered by a photoelectric cell.

THE LIGHT FLASH. Most scintillators tend to produce a single ultraviolet (UV) photon. Unfortunately, even materials transparent to visible light usually absorb ultraviolet light quite effectively, so most such UV photons are absorbed before they can be registered. The use of special impurities (about 0.1 percent) in scintillating crystals and solutions causes them to emit 2 or more low energy visible photons instead of a single UV photon. These "wavelength shifters" thus yield photons from the crystal, which are of a color more readily detected by both the eye and photoelectric cells.

For reliable detection, the received light flash should be as bright as possible. The brightness depends both on how much light is initially produced and how efficiently it is communicated out of the crystal. The original brightness is proportional to the energy transferred by the incident particle to the medium. For α -, β -, and soft x-rays this is almost all the available energy. High energy gamma and x-rays are much more penetrating, so large crystal thicknesses and high Z absorbers are often required for good photon absorption (such as NaI(Tl)). In addition, the phosphor efficiency of light production is also involved.

LIGHT COUPLING. Most of the light flash energy never reaches the photocell cathode. Consider a NaI(Tl) crystal next to a photomultiplier tube [Fig. 11(Left)]. Suppose two successive identical gamma rays are effective in producing light flashes, one at A, the other at B, closer to the tube. The dotted lines show directly transmitted light rays; it is evident that the light from B will be absorbed less than that from A, producing a larger signal. Were there a crystal imperfection in either path, the corresponding light received by the photocell would be greatly reduced.

Several pitfalls exist in getting the light to the photocell sensitive surface. First, much of the light is initially directed away from the photocell. Use of reflectors around the crystal helps reduce this loss. Another is absorption within the crystal, already mentioned. Finally, light can be reflected on the way to the photocell. If many small crystals are used, multiple reflection seriously reduces the signal [Fig. 11(Right)]. The use of small crystal units is limited to ZnS(Ag)

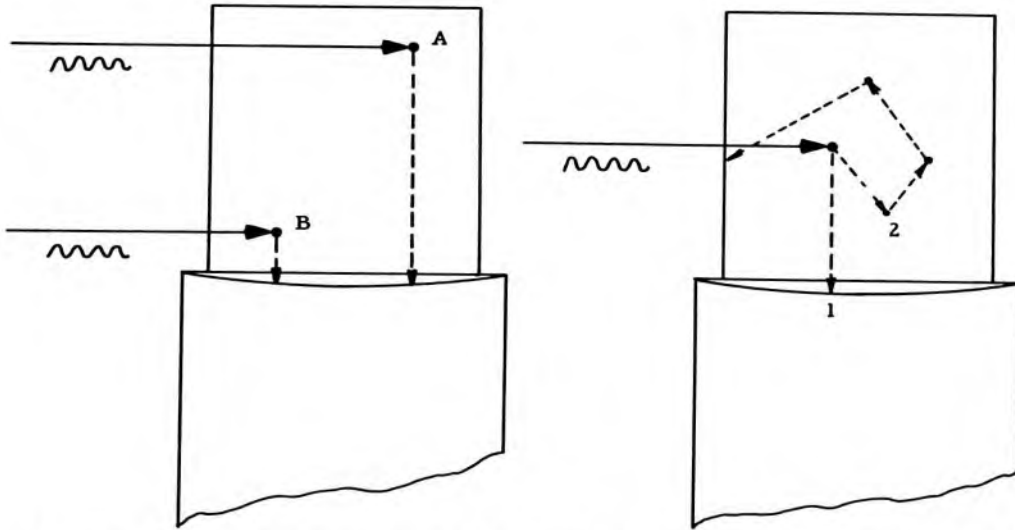


Fig. 11. Transfer of scintillator light flash to phototube. Left. Single NaI(Tl) crystal. Here the light (dashed lines) is communicated with minimum loss because the crystal is optically uniform. The crystal surfaces are made reflective to reflect light to the phototube which would otherwise be lost. A special sealing material is used between the phototube face and crystal to minimize reflection losses. Even an optically uniform crystal absorbs some light. Hence, for identical gamma rays absorbed at A and B, the electrical pulse from the B is slightly larger than from the A event. This is one of many causes of relatively poor resolution in scintillation systems. Right. Multiple crystals (details not shown for simplicity). There is a relatively small chance of a directly communicated ray (1) because of multiple reflections. Ray 2 indicates three such reflections. In general a single crystal must be used for both sensitivity and spectrometry reasons.

screens for α -ray or low energy x-ray detection. NaI(Tl) crystals for gamma detection are hence grown as large, single units, with a minimum of imperfections to reduce losses from internal reflections. Some crystals are 5 inches, 8 inches, and even greater in diameter.

Undesired reflection can occur at the junction of the crystal and the photocell. It is minimized by the use of good coupling optical greases of appropriate index of refraction. This optical coupling is critical to proper function; and any deterioration of this seal is disastrous to the signal and nullifies system reliability. *This criticalness of the optical coupling is a basic limitation to portability of scintillation systems.*

One more point should be noted regarding the light pulse. Since this is essentially a fluorescent signal, it can be released rapidly or slowly depending on the "phosphorescent" properties of the material. If the light energy is released quickly, a large pulse of short duration results; otherwise, the pulse is smaller and more prolonged for the

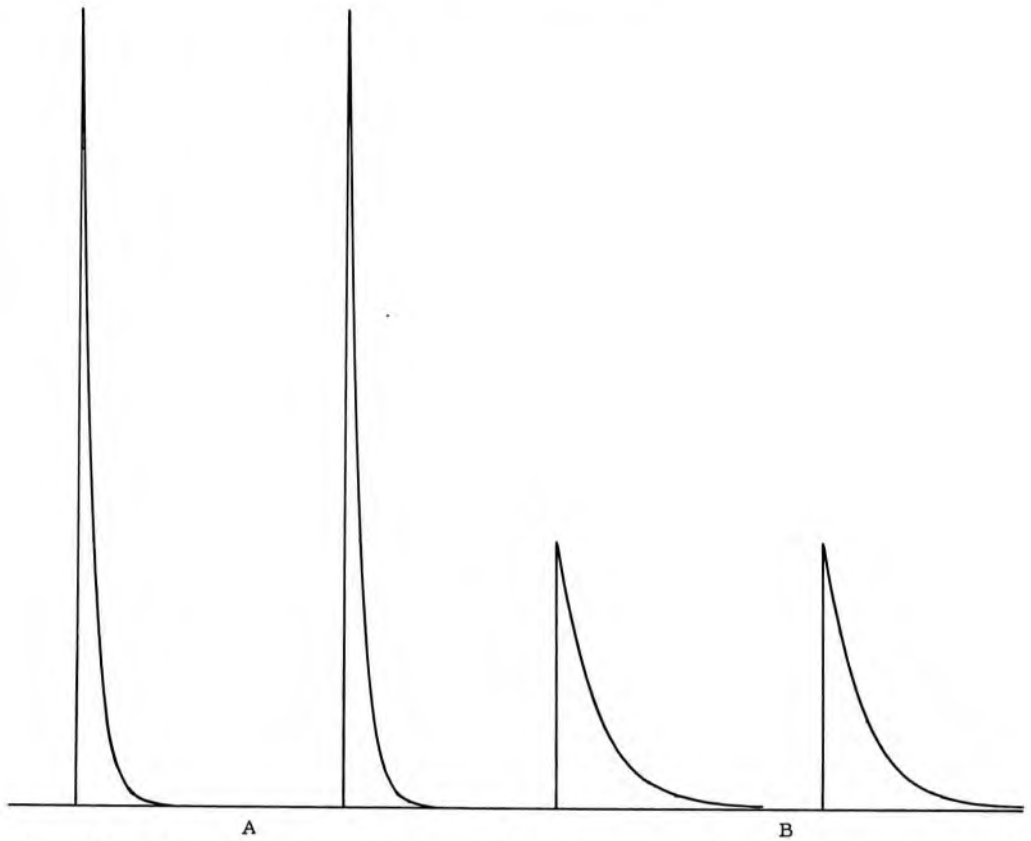


Fig. 12. Phosphorescence affects pulse size as well as duration. A. Normally pulses are of short duration. B. Some crystals protract emission of light considerably. Shown are two pulses of the same total light as in A. Note their protraction (phosphorescence) results in a corresponding reduction in pulse size, since the same energy is delivered for a longer time. As a result phosphorescence reduces pulse size as well as resolution.

same fluorescent energy (Fig. 12). Note that excessive phosphorescence acts like the G-M tube recovery time to limit resolving ability of scintillation counters. This effect, however, is usually unimportant in medical work, since the longest pulse decay times are of the order of 1 microsecond for inorganic crystals like NaI(Tl) and counting rates are kept low by patient safety considerations. Organic crystals respond more quickly, whereas organic liquid scintillators (such as terphenyl or diphenyloxazole in toluene or xylol) are up to 1,000 times faster.⁵

ELECTRIC PULSE PRODUCTION. After leaving the scintillator, the light pulse strikes the sensitive surface of a special photomultiplier tube. Although the engineering details of such tubes are not of primary interest here, their two basic functions are vitally important: to produce an electric pulse and then amplify it. Photomultipliers are photocells which amplify (i.e., multiply) their own currents by factors up to 10,000,000 times. Those used in scintillation counters generally have a very thin layer of cesium and antimony oxides at the end

facing the scintillator. The light readily releases photoelectrons from these particular materials. The photoelectrons are then attracted to a system of electrodes held at consecutively increasing positive potentials (in steps of 50 to 200 V). Most of the electrons, however, strike these electrodes without remaining on any but the last one. In fact, the electrode materials, shapes, locations, and voltages are arranged so that each impact of an electron releases several *additional* electrons, which then accompany the original electron to the next electrode. (These electrodes except for the last are usually called “dynodes.”) The net effect is a multiplication of the tube current pulse by about 2 to 5 times at each stage.

The output of the photomultiplier is thus an electric pulse whose size depends on both the energy initially absorbed by the crystal and the photomultiplier gain. The latter varies critically with the photomultiplier voltage employed which in turn determines the interdynode voltages.

Usually the main instrument is separated from the detector unit by a cable. Despite the amplifying action of the photomultiplier, its output signal is relatively low in energy. As a result, this signal would be severely distorted in shape by direct connection to the cable and instrument, so a preamplifier is provided (Fig. 13). This is usually a “cathode-follower” preamplifier, which actually reduces the pulse *voltage*

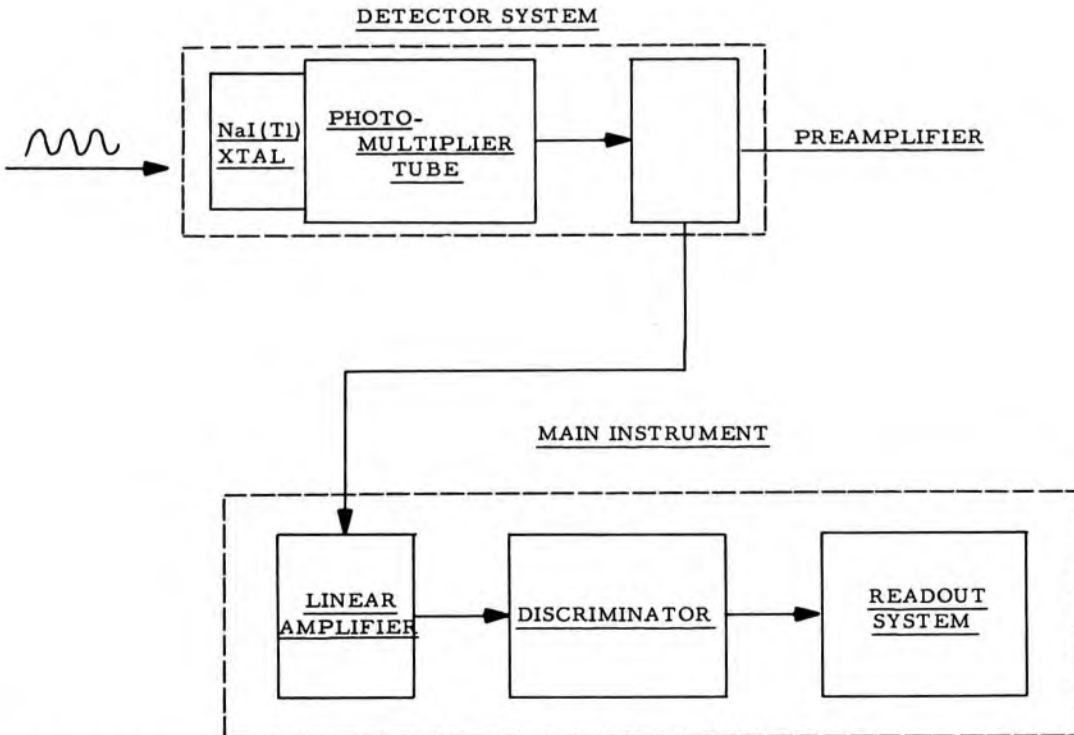


Fig. 13. Block diagram of electric pulse production and processing in a scintillation detector and instrument (see text).

slightly; it increases its *energy* greatly, however, preparing it for the trip to the main instrument. (More technically, the cathode follower serves as a coupling amplifier to match the impedance of the photomultiplier to that of the cable and instrument. If it is omitted, the electrical pulses are distorted with loss of information.) There, the pulses are first selected, then sorted, counted, and the result displayed (see below).

THE DETECTOR ASSEMBLY. It is evident that several cables must reach the detector assembly:

1. A special low-loss cable ("coaxial") to convey the pulses.
2. A high voltage cable. As indicated above, the high voltage must be extremely well stabilized or inconsistent pulse sizes (heights) result from the same initial energy absorption in the scintillator.
3. Supply voltages for the preamplifier.

Other requirements for the scintillator unit include:

1. A *light-tight seal* around all crystals, as extraneous light can greatly reduce the photomultiplier response to tiny scintillation signals.
2. *Magnetic shielding* required, as the photomultiplier amplification changes with nearby magnetic fields. (These deflect the electrons traveling between dynodes.)

Scintillation Detector Assemblies

As indicated above, ionizing radiation can produce light scintillation in both solids and liquids. Solid crystals are easiest to use, requiring only that they be shielded from light and attack by water and chemicals, a protection usually accomplished by sealing crystals in aluminum cases. They can then be used almost indefinitely without special preparation of samples. They can normally be used only to detect gamma rays, however, because beta rays are effectively absorbed by the aluminum case before reaching the crystal.

Liquid scintillators are used for counting samples emitting low energy beta rays. Samples are dispersed with an organic scintillator and other materials in a suitable solvent. This procedure is more costly in time and materials and is mainly of research interest at present. (See Chapter 11 for further discussion of liquid scintillation counting.)

The rest of our discussion of scintillation assemblies deals with crystal units only. Two types, probe and well counters, are discussed.

PROBE COUNTERS. Probe units are used for *in vivo* measurement external to the body. They can be kept fixed in location over an organ

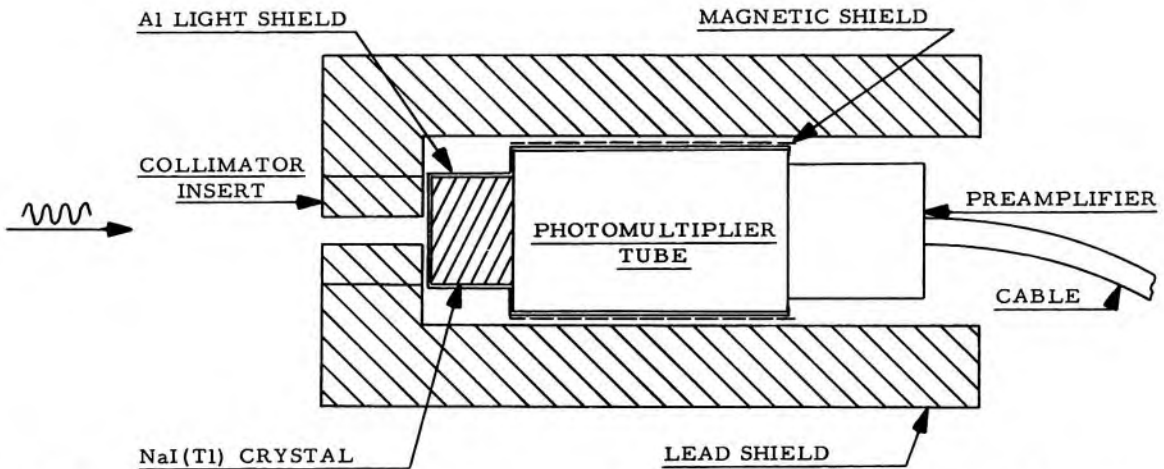


Fig. 14. Probe scintillation detection. Top. Thyroid uptake arrangement. Bottom. Essential parts of a scintillation detector probe.

for a single measurement, as in thyroid uptake studies, or moved over an area during scanning. Figure 14(Top) shows an ^{131}I uptake measurement setup and Figure 14(Bottom), the essential parts of the probe employed. Note the crystal detector, photomultiplier tube, and preamplifier units. An aluminum protective shield surrounds the crystal and photomultiplier tube. In addition, a magnetic shield and special optical coupling to the crystal are provided for the photomultiplier

tube, as described above. There are two additional components: a lead shield and a lead collimator insert. The shield absorbs many photons which would otherwise be detected coming from sources other than that being measured. Although thicker shields are more effective absorbers, positioning problems arise in using very heavy probes; in most cases, therefore, 0.5 inch, or at most 1 inch, lead thickness is used. The collimator is a device placed in front of the crystal to restrict its field of "vision" to the desired area. We shall have more to say about collimation in Chapter 11.

External detectors are of course inherently able to sample only a small fraction of the gamma rays emitted from the organ studied. This fraction can be increased by the use of larger crystals but only

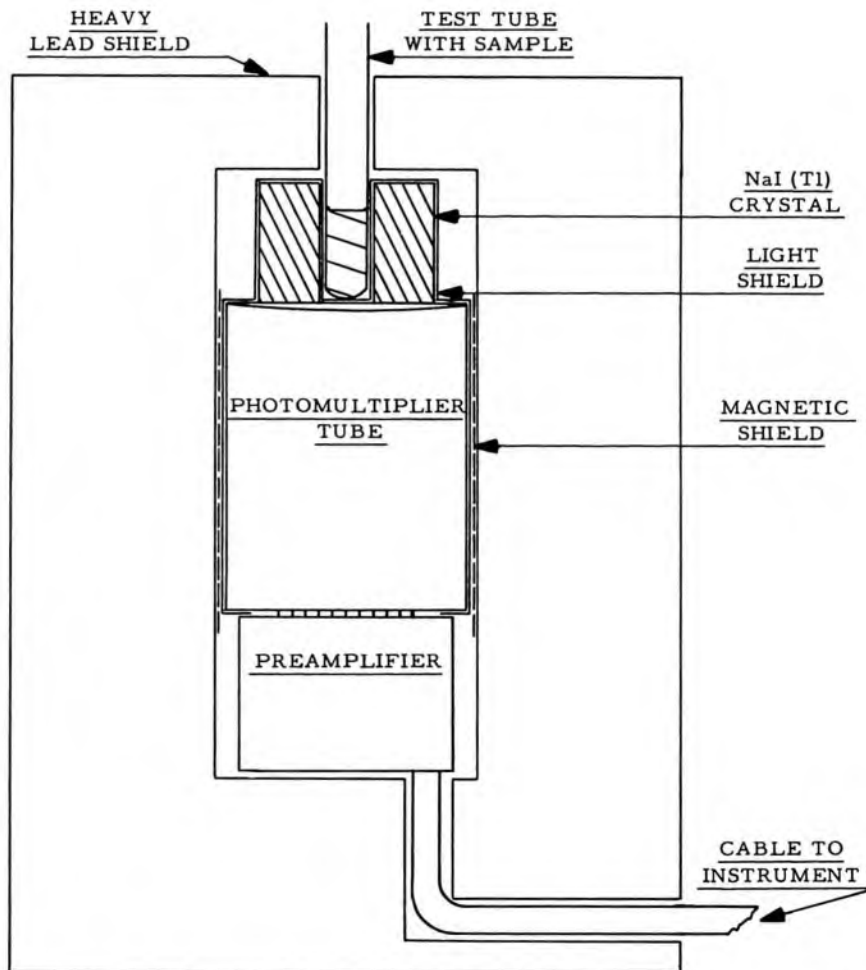


Fig. 15. Well scintillation counter essentials. The test tube containing the sample is surrounded by an annular NaI(Tl) crystal. This geometry greatly increases detection efficiency because most emitted photons strike the crystal. The rest of the system is essentially the same as that of a scintillation probe unit, with one major addition: much more lead shielding is used. This is permissible because well counters are used to measure samples only and are therefore fixed in position.

at considerable expense. Thus, geometry is inherently unfavorable in all probe measurements, so they are generally used only for in-vivo measurements, in which no better alternative exists.

WELL COUNTERS. Well counters have been developed specifically for very sensitive detection of radioactivity in samples which can fit into small test tubes. The detecting crystals are made annular in shape, and the sample placed in the central hole—*inside the crystal* (Fig. 15). Since the crystal substantially surrounds the sample most of the emitted gamma rays strike it and produce useful pulses. The rest of the system components are basically similar to those in a probe unit; however, usually more lead shielding is provided to improve counting statistics (see below) because the counter is stationary and weight is not a major problem. A good, adequately shielded well counter is extremely sensitive, and can conveniently and reliably detect less than $0.0001 \mu\text{Ci}$ of ^{131}I in routine clinical studies.

One caution is worth mentioning regarding radioactive contamination. Due to its shape a well is difficult to clean. Radioactivity trapped in it will add to all readings, effectively raising the background undesirably. Contamination can occur from leaky plastic sample test tubes, adhering radioactive material on test tube surfaces, etc. The problem is much less serious with short-lived than with long-lived radionuclides such as ^{60}Co or ^{137}Cs . A reasonable precaution is employing a finger cot around the end of the test tube whenever measuring long-lived radioisotopes. (Thin aluminum or brass removable insert shields for this purpose are also commercially available.)

Instrument Systems

A scintillation detector requires a rather elaborate associated instrument. Table 4 indicates the various functions of such instruments. They are of three general types:

1. To provide necessary voltages to operate the scintillation detector unit.
2. To process the pulses.
3. To display the result in useful form.

In more specialized instruments like scanners and automatic sample changers, there is a fourth function as well: to program all the steps in obtaining and recording the desired information. We shall, however, restrict our discussion here to the above three basic functions.

OPERATING VOLTAGES TO THE DETECTOR. The most critical voltage is that to the photomultiplier tube. This high voltage supply must be especially constant and reproducible because photomultiplier tube

TABLE 4. BASIC FUNCTIONS OF INSTRUMENTS USED WITH SCINTILLATION DETECTORS

A. Provide Supply Voltages for Detector Unit	
1. PM high voltage	very stable, conveniently adjustable.
2. Preamplifier voltages . . .	B+ generally, as well as filament, if vacuum tubes are used.
B. Process Pulses	
1. Linear amplification	about 1,000 times with great constancy and minimum change of shape and relative sizes of pulses.
2. Discrimination	select pulses, rejecting undesired signals.
3. Summate pulses	to yield count total and/or count rate.
C. Display Result	
1. Simple scaler	total on register, with elapsed time on timer.
2. Simple ratemeter	produces dc voltage proportional to c/m; displays result on meter or recorder.
3. Digital printout	prints time interval or total count at preset count or time intervals, respectively.
4. Computer	can process digital data to give desired display of result (curves, scans, etc.).
5. Scanning	graphic record produced to indicate distribution of radioactivity in an area.

gain changes greatly with small changes in voltage. For example, a 1 percent increase in voltage to a certain tube produces a 10 percent increase in pulse size. Since pulse size is used to select desired signals in discriminator circuits, extremely stable high voltage supplies are required for most scintillation counting work.

The requirements of preamplifier units are somewhat less severe, but their supply voltages must also be quite well stabilized.

PULSE PROCESSING. Pulses are handled in three steps: amplification, discrimination, and computation.

Since pulses from the detector unit are relatively small, they are first amplified greatly by a linear amplifier. In addition, the amount of amplification must be very accurately reproducible and constant, with minimum distortion of pulse shape. For rapid counting rates and detection of small pulses (as those from ^3H and ^{14}C beta rays in liquid scintillation counting) the demands on these amplifiers can be

especially exacting. For most medical gamma counting, however, the requirements are not unusually demanding.

Pulse "discrimination" refers to rejection of undesired pulses on the basis of their size, an ability which is one of the major advantages of scintillation counting. This subject is discussed in more detail below.

An instrument typically computes either the total count or the rate of counting, as indicated previously. Simple scalers and ratemeters are ordinarily adequate for most purposes. When ratemeters are unsuitable, fast printout systems are superior but may require considerable data plotting. (New digital computers and automatic plotters can perform this chore automatically.)

DISPLAY OF INFORMATION. Scalers and fast printout units provide numerical answers, usually in the form of total counts and elapsed times. Ratemeters compute the rate directly and display the answer on a meter or recorder. Scanners produce a visible record of the distribution of radioactivity over an area (Chap. 11).

PULSE HEIGHT DISCRIMINATION—USE. In clinical tracer studies one usually wishes to discriminate against four types of signals: natural background, scatter, photomultiplier noise, and, in certain two-tracer studies, those from another radionuclide.

Background radiation must be considered because very low amounts of tracer radioactivity are used in clinical work to minimize both somatic and gonadal irradiation of the patient. This correspondingly reduces measured counting rates to levels comparable with those produced by other sources, called "background" (from natural radioactivity, cosmic rays, and laboratory radioactivity and x-rays). Background sets a limit on the useful sensitivity of the instrument (p. 376).

Figure 16(Top left) shows the effect of scatter in a patient. We desire to measure radioactivity from an area B. If radioactivity is also present at A, the detector D receives not only the direct radiation (solid line) but also radiation scattered to the detector from the same area (dashed line). In many situations, the scatter pulse rate can exceed the desired one. Since photons are reduced in energy during Compton scatter, a system which rejects photons of lower energy can reject scatter and count the direct photons.

This principle is illustrated in the rest of Figure 16. Top right shows the spectrum of a monochromatic gamma emitter without scatter; that with scatter is shown at lower left. The actual corresponding counting rate is shown at lower right; the curve is somewhat rounded off because of the characteristics of detecting systems. Note many low energy scattered photons result from a few high

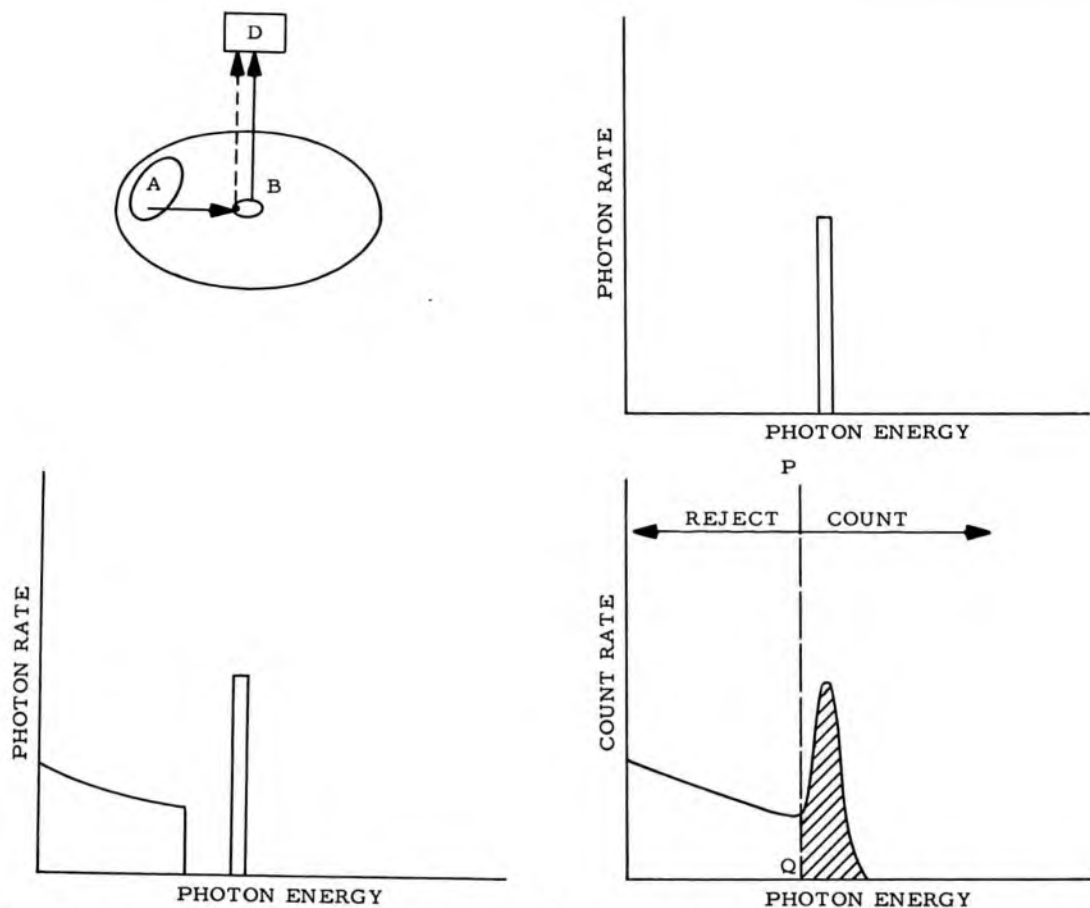


Fig. 16. Spectra of photons reaching detector during in-vivo measurements. Top left. *Scatter*. Both direct ray from B (solid line) and scattered ray originally from A reach detector from the same direction. Detector itself cannot separate them, so both produce pulses. Top right. *Spectrum*. Assumed single photon energy of direct radiation from B. Bottom left. *Spectrum*. Direct plus scatter radiation reaching detector. Bottom right. *Spectrum as measured*. Limited system pulse energy resolution distorts measured spectrum. Note the great contribution of scattered photons in bottom diagrams.

energy photons, so the problem could be serious if all were counted. Fortunately the discriminator can be set to reject all photons below line PQ, thereby counting primarily the desired photons. An incidental benefit is rejection of some low energy noise pulses. These originate as thermionic emission from the photoelectric surface and from traces of gas in the photomultiplier tube.

Finally, it is possible to evaluate the separate radioactivities of two different radionuclides in the same sample, with more elaborate pulse height discrimination (for example, ^{59}Fe and ^{51}Cr in hematologic studies). This is possible because they have different gamma-ray energies.

It is safe to say a great deal of the clinical usefulness of modern tracer techniques is based on the use of pulse height discrimination.

BASIC PRINCIPLE. How is pulse height discrimination accomplished? After amplification by the linear amplifier, pulses are made to run the gauntlet of a "channel." This is an electronic sorter which selects pulses corresponding to their sizes. For example, consider three pulses produced in a crystal by photons of 200, 360, and 1,000 keV. (These might be a scatter, ^{131}I , and radium photon, respectively.) In a particular, very efficient detector these might result in electrical pulses sent to the amplifier of 20, 36, and 100 mV pulse amplitudes. After 1,000-fold amplification the channel would then be confronted with three pulses in sequence of 20, 36, and 100 volts. Suppose this channel were set to reject all signals smaller than 32 and larger than 40 volts (corresponding to 320 and 400 keV). Then, only the desired 364 keV ^{131}I photon would be registered; pulses from both the scatter and radium rays would be rejected. Two pulse height discriminators may be used with each channel, one to reject low, the other high pulses, so only the intermediate ones are counted.

SPECTROMETERS. Many degrees of pulse processing sophistication exist in useful systems. The simplest has only a low energy cutoff, which is not adjustable. This could be factory-set to reject ^{131}I scattered rays but would be unsuitable for others. More versatile is an adjustable version, to permit setting the acceptable lower energy value at will in accordance with the radionuclide employed and other variables. In both these simple units, however, there is no rejection of higher energy radiation.

The next step is a simple "spectrometer," which adds a *high* photon energy cutoff. In addition, it normally provides precision adjustments and precision stabilized electronics, so the two discriminators can be accurately and reproducibly set. Naturally these require more expensive circuitry and components. Since there is only a single channel available at any one time to select pulses, this is a single-channel spectrometer. The allowed pulse height width is often called the channel "window," and the low cutoff value is the "window sill."

For example, in our previous example the window was $400 - 320 = 80$ keV, and the window sill was 320 keV. Both the window and the window sill are adjustable in this instrument.

Single-channel analyzers are usually sufficient for most clinical work. In fact, their very adjustability is useful in those areas of clinical practice in which different radionuclides are used.

In some specialized work, it is desired to automatically display an actual count rate versus photon energy spectrum of the photons reaching the detector crystal. One could use a single-channel unit and count sequentially, using narrow windows and gradually raising the sill stepwise (for example, in steps of 6 keV, with corresponding "sills" of 6, 12, 18, 24, etc.) This would require 250 counts to reach 1.5 MeV—a very laborious procedure. Special instruments have therefore been built with many channels to do this automatically. Each pulse is simultaneously tested on up to 250 channels and passed by the right one. The individual totals are stored and later displayed appropriately as a graph on an oscilloscope, printed on tape, etc. Such an instrument is called a multichannel analyzer.

PRACTICAL SELECTION OF EQUIPMENT. One is often asked, "What is needed as basic equipment to get started in nuclear medicine?" Although the answer of course varies greatly with circumstances, Table 5 indicates some common types of equipment and the uses to which they may be put. It must be emphasized these instruments are

TABLE 5. SOME NUCLEAR MEDICAL INSTRUMENTS

A. Measuring Instruments and Equipment

1. **Basic ^{131}I uptake setup***
 - (A) Scintillation probe with shield.
 - (B) Scaler with very stable HV supply, good timer.
 - (C) Discrimination, adjustable upper and lower cutoff.
 - (D) Thyroid phantom reference unit.
2. **Small portable scintillation probe and scaler for placentography, etc.**
3. **Scanner:*** 3" crystal with photoscanning, 19-hole focusing collimator. (*Options:* 5" crystal vs. 3", special collimators, color or digital scanning, two to four scanning heads, and special newer cameras).
4. **Dual-channel ratemeter for kidney function studies, etc.**
Option: digital printout, for faster response studies.
5. **Well counter and spectrometer.***
 - (A) Two inch vs. one inch Pb shielding.
 - (B) Super-stable HV supply.
 - (C) Simple spectrometer unit.

B. Radiation Safety

1. Small portable thin window G-M counter unit.*
2. Cutie pie, logarithmic scale.*
3. Well-type ionization chamber meter system for radionuclide shipment assay.

* Basic units for common studies and general use.

not inexpensive, and a nuclear medicine laboratory capital investment of \$25,000 to \$50,000 may be easily reached without extravagance. Items marked with an asterisk (*) are required for the most common studies.

Liquid Scintillation Counting

We have previously discussed problems of detecting very soft beta rays, such as those from ^3H and ^{14}C , and indicated the superiority of liquid scintillation counting for such work. Liquid scintillation counting involves rather sophisticated technology, however, so the present brief discussion has been deferred deliberately. Several aspects are of basic interest.

The sample is specially prepared. A clear and preferably colorless solution is made of the scintillant liquid preparation dissolved in an aromatic hydrocarbon solvent like toluene. Although lipids present no solubility problem, water-soluble materials often require special dispersion to obtain accurate results, and proper standard preparation techniques must be employed. In addition to the actual scintillant and sample, two other materials are usually required. The first is an activator to derive longer wavelength photons from the short wavelength ultraviolet photon which is directly produced by the initial excitation. The second is an antifreeze material to prevent solidification of the sample because the entire system is often refrigerated for reasons discussed below. The sequence starts with absorption of the beta ray by the solution. This excites the scintillator and activator, producing a very faint light flash of brightness proportional to the absorbed beta ray energy. For ^3H these are extremely faint signals.

Special light detection systems have been developed because photomultiplier tube noise signals are comparable in magnitude to photoelectric pulse signals from ^3H . This problem is customarily solved by use of a combination of photomultiplier tube selection and deep freeze operation. In addition, *two* photomultipliers detect the pulse simultaneously. When a light flash occurs, it strikes the cathodes of both tubes, producing *simultaneous* pulse signals. These differ in this regard from the photomultiplier tube noise signals which are *random* in timing. Special "coincidence" circuits count only the pulses produced simultaneously in both PM tubes, rejecting the others; this excludes most noise pulses from the counting circuits.

Fairly recently Beckman developed a new very low noise photomultiplier tube in conjunction with one of the vacuum tube companies. It is claimed that this tube does not require refrigeration, thereby greatly reducing associated practical difficulties and cost.

In addition, the photocell response extends further into the ultraviolet than that of older tubes, so the wavelength shifter solution is not required. This instrument costs less as a result, and the "scintillation cocktail" is much simpler and less expensive to prepare. It will be interesting to see how this system works in practice in the coming years.

The basic advantage of liquid scintillation counting is its great sensitivity and potential accuracy at low beta ray energies. Modern units, in addition, are capable of counting 100 to 200 samples automatically, a great advantage in many research activities. Generally, samples can be prepared during the day and loaded, then counted automatically overnight.

Liquid scintillation systems are not inexpensive (\$8,000 to \$15,000, typically). Solvent and scintillant, as well as other materials, are also not ordinarily reusable, adding to the cost of each measurement. Liquid scintillation is necessary, however, for tritium counting and very convenient in other beta counting work if large numbers of samples are measured.

MEASUREMENT ACCURACY

Various instruments for measuring radioactivity have been described. Many pitfalls exist in using these instruments for specific applications, however, and serious errors await the unwary. Some of the major pitfalls, including three different types, are discussed in some detail in this section:

1. *System errors*, arising from faulty equipment and procedures.
2. *Statistical uncertainty errors*, inherent in all low level radioactivity counting.
3. *Dynamic errors*, arising mainly in measuring rapidly changing radioactivity.

System Errors

It is instructive to recall the simple problem of measuring the length of a table. Two kinds of errors may arise in doing this. First, an inaccurate yardstick with an obviously erroneous scale may be used. Second, the length may be measured improperly by setting the zero end incorrectly or inclining the yardstick. In the first case, the problem is faulty equipment or instrumentation; in the second, faulty procedure (due perhaps to poor *mental* equipment). Such errors of faulty equipment and procedure arise in what might be called the overall measuring system and are hence called system errors.

TABLE 6. COMMON PITFALLS IN RADIATION MEASURING EQUIPMENT

A. Radiation Detectors

1. G-M tubes

- (A) Plateau change with aging and use.
- (B) Resolving time errors, causing low counting rate indication.

2. Scintillation detectors

- (A) Optical coupling changes between crystal and PM tube.
- (B) PM tube noise; effects of strong magnetic fields.
- (C) High voltage supply variation with aging, warmup.

3. Any type

- (A) Aging produces calibration changes; regular check needed.
- (B) Cable connectors: spurious pulses or lost counts are possible.

B. Instruments

1. Any type: scaler failure, caused by

- (A) Bad glow tubes and mechanical registers.
- (B) Other scaler parts (tubes, transistors, capacitors, etc.).

2. Changes in discriminator settings and amplifier gain.

3. Ratemeters: inherent inaccuracy of ratemeter circuit and indicating meters.

EQUIPMENT PROBLEMS. Table 6 summarizes some common equipment problems which have been observed in radioactivity measurements. Many of these are obvious, but others require comment.

All radiation detectors change in sensitivity with time. It is therefore necessary to use either absolute or reference standards to determine the detector sensitivity *when used*. Usually, a relative standard can be made up conveniently along with the administered patient dose.

A common culprit is a faulty high voltage cable. It is readily checked by substituting a spare cable for the suspect one when instruments act erratically. This simple procedure can often avoid unnecessary shutdown delays.

Any scaler circuit will ultimately require service; trouble is usually indicated by intermittent "skipped" counts (yielding too few measured counts). A good technician constantly checks the consistency of his measured data. Also, he grows accustomed to the rhythmic progression of glow tube lights and almost unconsciously notes skipped counts. Alertness in this regard can greatly minimize incorrect or repeat studies from this cause towards the end of scaler glow tube life.

A good quality spectrometer unit usually has very high stability of both discriminator voltage and amplifier gain; however, one should keep this requirement in mind in purchasing such instruments.

TABLE 7. SOME PROCEDURAL MEASUREMENT ERRORS

A. Examples

1. **Background change during measurement; contamination of instrument.**
2. **Geometry, absorption, and scatter not reproduced.**
 - (A) G-M beta counting: variation in sample preparation, holders, planchets and distance.
 - (B) In-vivo scintillation probe counting: phantom patient used with standard must simulate actual patient attenuation and scatter.
 - (C) Well counting: requires constant test tube type and sample volume.
3. **Dial "twiddling"**—discriminator or HV supply settings altered accidentally.
4. **Computational errors.**
 - (A) Division: computation of c/m values.
 - (B) Subtraction: background correction.
5. **Gas flow counting:** contamination, improper gas flushing, breakdown of collector insulation (spurious counts).

B. Recommendations

1. **Well designed basic study procedure with suitable equipment.**
2. **Consistent checking and critical evaluation of results.**
3. **Corridor radioactivity alarm to warn of approach of high level activity such as radium applicators, shipments, radioactive patients, etc.**
4. **Procedures**
 - (A) Background: run 4 times daily. Check trends daily.
 - (B) Sensitivity: check with calibrated sources. Note trends.
 - (C) Plateaus: regular checks.
 - (D) G-M tubes: for higher counting rates, minimize resolution errors by using reference of activity reasonably comparable to that of samples.

PROCEDURAL PROBLEMS. Table 7 summarizes some common procedural problems with some general recommendations.

Background is the measured count rate without the patient or sample in place. It can vary during measurement of the patient, sample, and standard, or in the interval between these measurements. Some possible causes to keep in mind are: radium removed from the safe or moved about the department in a lead carrier or patient, opening of a shipment containing a high activity gamma emitter, and the nearby operation of a supervoltage machine whose shielding is marginal. Normally, well counters can be very well shielded; use of an extra 2-inch Pb shield is often desirable. It is unfortunately more diffi-

cult to shield probes because they must be light enough for easy maneuvering over the patient.

Exact reproduction of patient geometry, scatter, and absorption in phantom measurements is generally quite difficult. Fortunately, clinically useful results are often achieved with fairly simple arrangements, and standardization of procedures at least assures reproducibility of procedures among various laboratories. For example, this has been done in thyroid uptake work.

Most other items of Table 7 are self-explanatory. Some items of B-4, however, merit special comment. Running background counts for suitable periods four times daily (twice each morning and afternoon) is likely to detect most background fluctuations in time to conveniently repeat measurements when necessary. Sensitivity checks are best carried out using known activities of a standard reference source. A convenient source for constancy checks is a 5 or 10 mg radium capsule placed at a fixed location relative to the detector, with reasonably good reproducibility of scatter and attenuation conditions. Another inexpensive and more convenient source is a ^{60}Co or ^{137}Cs solution of suitable activity.

G-M tube plateaus should be checked at least monthly, more often if frequently used. This check is done by measuring the count rate of a beta or gamma source of suitable constant activity in a fixed position while gradually increasing the G-M tube voltage. Of course, care must be exercised to avoid excessive tube voltage with attendant tube damage.

Statistical Uncertainty Errors

Radioactive decay is a random process. For example, although one can say that roughly half of a billion ^{131}I atoms will decay during its half life of 8.1 days, one cannot say exactly when any *particular* atom or group of atoms will decay. There is a chance, although very small, that at least 1,000 of these atoms will not decay for 10 *years*—long after their scheduled demise! The exponential decay law becomes less and less accurate in predicting the number of d/s as the activity decreases.⁶

This fact has great practical significance in clinical tracer work, in which considerations of patient dosage severely limit the activity employed. The result is that the measured count rate of a fixed activity is a variable rather than fixed quantity. The degree of fluctuation is small for large number of counts, great for small numbers.

Since we normally make a given radioactivity measurement only once, the result is generally too high or too low. For example, consider

TABLE 8. SUCCESSIVE MEASUREMENTS OF THE SAME ^{51}Cr SAMPLE—
COMPARISON OF FLUCTUATIONS IN LARGE
VS. SMALL TOTAL COUNTS

60 Seconds Counting Time			1 Second Counting Time		
Measured Total	Difference ϵ	ϵ^2	Total	Difference ϵ	ϵ^2
3,343	32	1,024	55	3	9
3,573	98	9,604	62	3	16
3,442	33	1,089	56	2	4
3,577	102	10,404	55	3	9
3,553	78	6,084	54	4	16
3,507	32	1,024	59	1	1
3,342	133	17,689	57	1	1
3,472	3	9	67	9	81
3,359	116	13,456	62	4	16
3,519	44	1,936	53	5	25
3,437	38	1,444	55	3	9
3,491	16	256	44	14	196
3,542	67	4,489	64	6	36
3,444	31	961	62	4	16
3,404	71	5,041	59	1	1
3,408	67	4,489	56	2	4
3,558	83	6,889	47	11	121
3,366	109	11,881	57	1	1
3,516	41	1,681	58	0	0
3,542	67	4,489	77	19	361

Average = $(3,475 \pm 72)$ c/m (2%),
or (58 ± 1.2) c/s

Average = (58 ± 6.8) c/s (12%)

Note there is a $\pm 12\%$ vs. $\pm 2\%$ variation in the counts measured for 1 second vs. 60 seconds, respectively.

a blood sample containing ^{51}Cr removed in the course of a blood volume study. For a particular sample this yielded 3,475 counts in 60 seconds, or about 58 counts per second. A series of one minute counts on this sample gave the values shown in the left column of Table 8. Although this total fluctuates a bit (about 2 percent on the average), it is relatively constant. Note by contrast the large fluctuations of successive *one second* counts (12 percent on the average). The lesson is plain: *to reduce statistical fluctuations, obtain a greater total count, either by using more activity or counting longer.*

In practical measurements a variety of problems arises involving statistics. There are, however, three questions of most frequent interest:

1. Exactly how large an error is likely in a *single* observation?
2. What happens to errors when readings are combined, as in correcting for background?
3. Finally, how can one check a set of successive readings to verify that *system* errors are not excessive?

UNCERTAINTY OF A SINGLE OBSERVATION. Early in the study of probability it was found that the more random events one counts, the more consistent his results. For example, there is one chance in sixteen a fair coin will turn up heads every time in *four* tosses; this is fairly good odds for a gambler. However, if one tosses the coin *12 times*, the odds in favor of having all 12 heads drop drastically to only one in 4,096—a very poor gamble. It compares with an even chance (one in two) of *equal* numbers of heads and tails.

In radioactivity, we are gambling on the exact time of disintegration of atoms rather than heads or tails of a tossed coin. It is the total detected counts rather than the number of coin tosses which determine how close we come to the theoretical counting rate predicted by the numbers of atoms present.

Statistical theory indicates the standard deviation of an observation of N counts should be:

$$\sigma = \sqrt{N} \quad (10-2)$$

For example, if 10,000 counts are registered on our instrument, the standard deviation (often loosely called “error”) from a true value is $\sqrt{10,000}$ or 100 counts. The standard deviation can be a very helpful concept. In general there is a 67.45 percent chance (about $\frac{2}{3}$) that *any* observation of N counts will be within σ or \sqrt{N} of the true value, a 95 percent chance of being within 2σ , and 99 percent chance of being within 2.6σ .

Usually we are less interested in the magnitude of σ than in its size relative to N . We hence define a second term V , the percent standard variation.

$$V = \sigma/N \times 100\% = \frac{1}{\sqrt{N}} \times 100\% \quad (10-3)$$

Table 9 illustrates how σ and V vary with counting time for a weak sample (counting rate = 500 c/m). It is evident that prolonged counting periods may be needed sometimes to increase the accuracy of measurement.

TABLE 9. COUNTING ACCURACY VS. COUNTING TIME—
500 c/m

Counting Time	Total Counts—N	Standard Deviation— σ	Percent Standard Deviation
1 min	500	22	4.5%
5 min	2,500	50	2.0%
20 min	10,000	100	1.0%

THE EFFECT OF BACKGROUND. It will be recalled that background produces detector counts even when the measured sample is removed completely. Background radiation arises from both natural and man-made sources. Natural signals result mainly from disintegration of radionuclides in the ^{238}U , ^{235}U , and ^{232}Th series, as well as from ^{40}K and ^{14}C . ^{14}C is formed continually from cosmic ray bombardment of ^{14}N in air; all the others are believed to be residual from materials present when the earth was formed. Direct ionization by cosmic rays also contributes to background but normally to a much smaller extent. Man-made sources include x-rays and stored radioactive materials and, to a lesser extent at present, fallout from nuclear explosions.

The detector inevitably responds to signals from both the sample and background. Hence, the measured *total* count rate R_T is the sum of the desired count rate R_O for the sample alone and the background count rate R_B , or, algebraically:

$$R_O = R_T - R_B \quad (10-4)$$

In general $R = N/t$, where R is the count rate and N and t are the corresponding count total and counting time, respectively. This common sense result gives us a simple procedure for getting the desired sample count rate from rates that can be measured.

When R_B is very small, its fluctuations can be ignored as they do not significantly affect the statistical fluctuations. Unfortunately, low radioactivity values in medical tracer work make R_B a significant fraction of R_T so it greatly degrades the statistics. Let us compute the effect of R_B on the statistical uncertainty of the final result.

Since $R_O = R_T - R_B$, basic probability theory tells us their corresponding σ values have the following relationship:

$$\sigma_{R_O} = \sqrt{\sigma_{R_T}^2 + \sigma_{R_B}^2} \quad (10-5)$$

In general the σ_R quantities are simply the σ quantities for the N 's divided by the corresponding times t . Simple substitutions lead to the useful final formula:

$$\sigma_{R_o} = \sqrt{R_T/t_T + R_B/t_B} \quad (10-6)$$

Since all the terms on the right are directly measured, σ_{R_o} is easily computed.

Two practical applications will serve to illustrate this useful formula.

I. ^{131}I uptake study.

Assume: 1,800 c/m total, measured for 3 minutes.

300 c/m background count, 10 minute count.

Find the statistical error.

a. $R_o = R_T - R_B = 1,800 - 300 = 1,500$ c/m.

b. Substituting in (10-5):

$$\sigma_{R_o} = \sqrt{\frac{1,800}{3} + \frac{300}{10}} = \sqrt{630} = 25 \text{ c/m (approx.)}$$

$$\therefore R_o = (1,500 \pm 25) \text{ c/m and } V_{R_o} = \frac{25}{1,500} \times 100\% = 1.7\%$$

c. Conclusion: In any particular measurement, the odds are:

2:1 of 1.7 percent or smaller statistical error

20:1 of 3.5 percent or smaller statistical error

100:1 of 4.5 percent or smaller statistical error

d. Note the background time could have been reduced considerably (e.g., to 2 minutes) without adding significantly to the over-all error while speeding up the procedure considerably.

II. Very low ^{131}I pickup in myxedema.

Take: 600 c/m total, measured for 3 minutes.

300 c/m background count, 3 minute count.

a. $R_o = R_T - R_B = 600 - 300 = 300$ c/m.

b. Substituting in (10-5):

$$\sigma_{R_o} = \sqrt{\frac{600}{3} + \frac{300}{3}} = \sqrt{300} = 17 \text{ c/m, (approx.)}$$

$$\therefore R_o = (300 \pm 17) \text{ c/m, and } V_{R_o} = \frac{17}{300} \times 100\% = 6\% \text{ (approx.)}$$

c. Conclusion: In any particular measurement, the odds are:

2:1 of 6 percent or smaller statistical error

20:1 of 12 percent or smaller statistical error

100:1 of 16 percent or smaller statistical error

d. Were there no background here, R_T would be 300 c/m and σ_R would be only 10 c/m. V would then be only 3 percent or half as great. It is evident background here doubles the statistical error.

TABLE 10. EXAMPLE OF CHI-SQUARE (χ^2) CALCULATION*⁵¹Cr sample measured 20 times for 10 seconds each time.

Total Counts N	Difference ϵ	ϵ^2
606	20	400
598	12	144
585	1	1
578	8	64
587	1	1
597	11	121
606	20	400
596	10	100
535	51	2,601
386	0	0
<hr/>		
578	8	64
568	18	324
606	20	400
576	10	100
591	5	25
606	20	400
582	4	16
570	16	256
585	1	1
574	12	144

Average 586

 ϵ^2 Total: 5,562 (Ave. = 278)

$$\chi_{20}^2 = \text{Sum of } \epsilon^2 / \text{average count} = 5,562 / 586 = 9.5$$

*From *Handbook of Chemistry and Physics*, 37th ed.,⁽⁷⁾ page 224. For 20 degrees of freedom (counts), p is between 0.98 and 0.95, indicating satisfactory degree of randomness.

THE CHI-SQUARE TEST. Both system and statistical factors can introduce measurement errors; however, only statistical errors stem from purely random events. Usually system errors show some kind of trend. The chi-square test involves repeated counts of a constant radioactivity source to detect any such non-random trend.

The basic procedure is briefly described below (see Table 10 for an example).

1. Take a succession of about 20 counts with equal counting time intervals. Average the data.
2. Compute the differences between the measured counts and the average value. Square the differences and summate these squares.
3. This sum divided by the average count is χ^2 (chi square).
4. χ^2 tables are given in various references (Table 10).⁷ Compare the χ^2 values from the data with those in the table.

5. If χ^2 is either too large or too small, the probability is that the distribution of numbers is not following a consistent pattern. Fluctuations are then not random, and a system error should be actively sought.

Dynamic Errors

We have thus far considered sources of error common to all radioactivity measurements: faulty equipment, procedures, and statistics. In most work these present the main pitfalls.

If the radioactivity changes rapidly, however, additional difficulties may arise in accurately recording these changes. If the counting is too slow, crude average values which do not represent the actual radioactivity result. Two kinds of situations arise in tracer work when such dynamic errors arise. The first is in measurements of living systems, in which physiologic and biochemical processes occur with considerable speed, and measured tracer quantities, therefore, also change rapidly. The second situation involves tracer elements available as only very short half-life radionuclides; these may decay significantly while being measured.

A full discussion of techniques for carrying out these more sophisticated measurements is beyond the scope of this book. The basic principles and pitfalls of such studies must be appreciated, however, to avoid significant errors in even more conventional studies. It is evident that if the radioactivity changes significantly during a minute or two, an ordinary scaler measurement will simply give a crude average indication [Fig. 17(Top)]. A similar type of erroneous indication results when a slowly responding ratemeter is employed.

Faced with this unsatisfactory result, one might employ a rapidly responding ratemeter with a fast recorder, which can respond to the instantaneous counting rate fairly accurately. Unfortunately the result [Fig. 17(Bottom)] also reflects large statistical fluctuations as well. Why should these be so large? Fast counting means sampling for only short periods of time; since time intervals are short, so are the measured counts, particularly in medical work where activities must be kept low.

There is thus a basic conflict: for good statistics one needs adequate counts; for dynamic accuracy, only a limited sampling time is available. With ordinary radionuclides moderately long half-lives and beta-ray emission impose severe limits on permissible administered radioactivity because of patient dose considerations.

Useful approaches to this dilemma probably all involve the following basic principles:

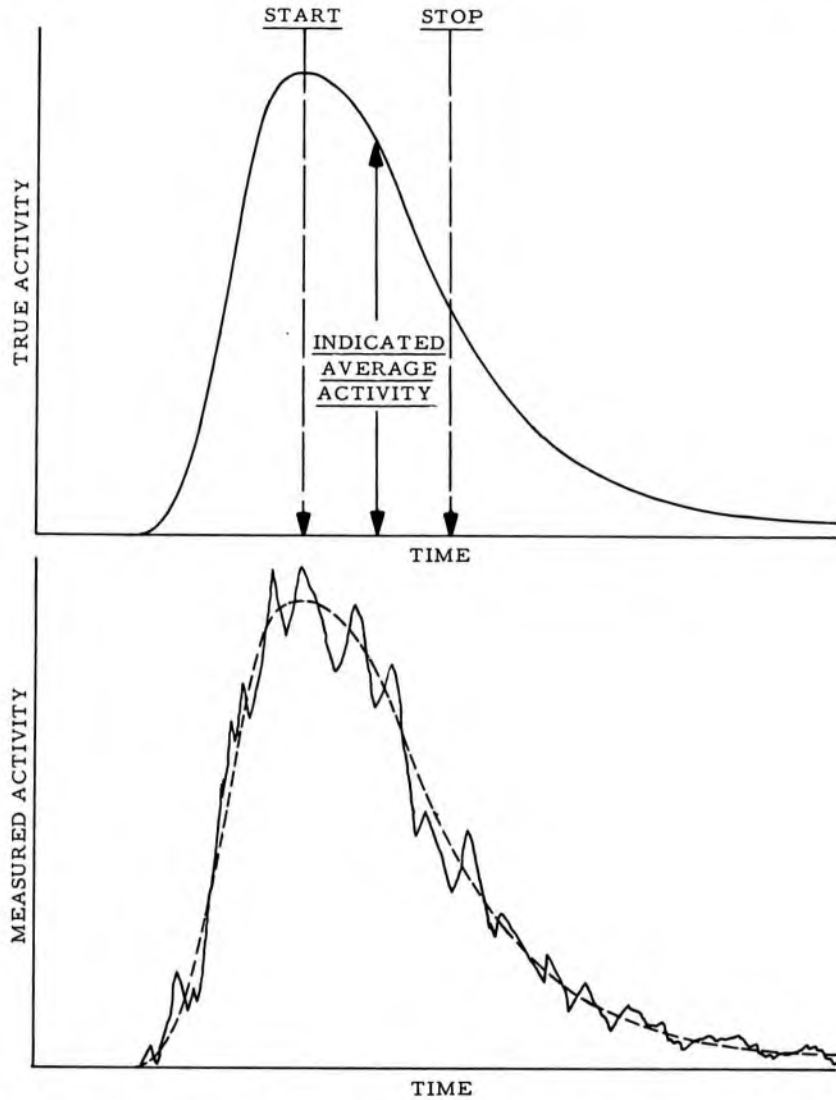


Fig. 17. Problems of measuring rapidly changing amounts of radioactivity. Top. Scaler limitation. For rapidly changing functions, activity can vary greatly during a convenient measuring time, and the result is a crude average value only. Bottom. Fast recording: solid curve—measured, dashed—true. In this case, statistics are unfavorable because activity is very low. Use of short-lived pure gamma emitters like $^{199\text{m}}\text{Tc}$ permits administering relatively high microcurie amounts without excessive patient irradiation. Such higher activities help greatly in reducing statistical fluctuation.

1. *Use of pure gamma-emitting and short half-life radionuclides for very rapid studies.* With such materials, patient dosage per administered microcurie is greatly reduced (as will be shown in Chapter 12), so one may safely measure much greater amounts of radioactivity, thereby greatly benefiting statistics.
2. *Use of very sensitive detectors.* This involves larger diameter

and thickness scintillation crystals to benefit geometry and efficiency.

3. *Use of rapid electronics and digital printout and computer systems.* These can provide rate indications with minimum pulse processing and display errors.
4. *Finally, use of some mathematics to analyze the results.* In some cases good planning can result in a relatively simple procedure for inferring useful conclusions directly. The use of available mathematics and physics guidance in planning difficult measurements can pay good dividends.

All these approaches offer promise. Of them all, the first is perhaps most basic and is already being implemented. Electron capture and isomeric transition radioisotopes of short half-life are hence becoming more widely employed. These and allied aspects of the problems are considered further in the next chapters.

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In-Vivo Radioactivity Measurement

Clinical tracer studies can employ two types of measurement. In the first, samples are taken from the patient at appropriate intervals after administration of radioactive material; these may be of blood and other fluids, excreta, and actual tissue samples. In the second, measurements are made essentially externally to the body, without removal of any material from the patient. We refer here to the former as "sample," the latter as "in-vivo" types of measurements.

In-vivo studies are generally preferred to alternative sample techniques for several reasons. First, many practical problems arise in obtaining samples. Multiple blood sampling is inconvenient and often hard to accomplish, particularly in older patients, and obtaining organ samples such as those of the liver or bone marrow may involve painful and even hazardous procedures. Second, samples provide information regarding only one local site at one particular time. Finally, the very act of sampling by knife or even needle is traumatic, and blood and other tissue accompanies the desired sample. The measured activity per gram of sample is hence generally a crude average unless special preparation is performed. Consequently, although excreta and blood samples are frequently measured, tissue samples are generally limited to carefully removed surgical specimens.

External gamma counting is virtually non-traumatic, resembling x-ray studies in this respect. It can be used to study organ function and body fluid flow by sequential or continuous measurements using a fixed probe. In addition, scanning techniques provide information about the distribution of radioactivity in and around an organ of interest, such as the brain, thyroid, liver, and kidneys. When repeated sequentially, scans reveal information concerning organ function and blood flow; this is similar in principle to sequential radiographic studies, even though the method is different.

In addition to fixed probe and scanning gamma counting, in-vivo beta and whole body counting are two other studies less frequently performed.

This chapter considers all these in-vivo measurements, in the following order:

1. Principles of probe measurement
2. Scanning
3. In-vivo beta- and whole body gamma-ray counting.

PROBE MEASUREMENT

Recall that external probes are generally shielded scintillation detectors, provided with appropriate collimators. Probes are used for both fixed and scanning measurements. Unfortunately, radioactivity is generally present in areas adjacent to those of interest. To some extent a probe can selectively measure the desired signals. Three variables all affect the degree of selectivity of a probe measurement: collimation, positioning, and depth in the body of the site of interest. These will now be considered briefly.

Collimation and Positioning

As used in radioactivity measurements, "collimation" is an elegant term for shielding used to selectively emphasize signals from a desired organ. It is not easy to do effectively since both the desired and undesired signals have the same photon energy and may come from the same general area.

WIDE-ANGLE COLLIMATION. In some measurements one desires the activity of an entire organ, such as the thyroid gland or a kidney. It is important in such a measurement to have nearly equal detector response to signals originating in the center and edge of the organ. Often, a relatively large area is sampled because the full extent of the organ is not known. Wide-angle collimation, which excludes a minimum of direct organ radiation from the crystal, is employed for this purpose. In addition, a fairly large detector-organ distance (usually 20 to 30 cm) is used to achieve reasonably uniform response to radiation from all parts of the organ. In general, one selects an arrangement of collimation and distance which adequately excludes the major sources of interference while providing good response to photons originating in the organ.

Thyroid uptake measurements are the most important study of this type [Fig. 1(left)]. Note the entire thyroid is usually "seen" by the detector, but the heart and brain blood pool radiation is effectively shielded. Other sources in the neck however cannot be shielded without loss of thyroid signals.

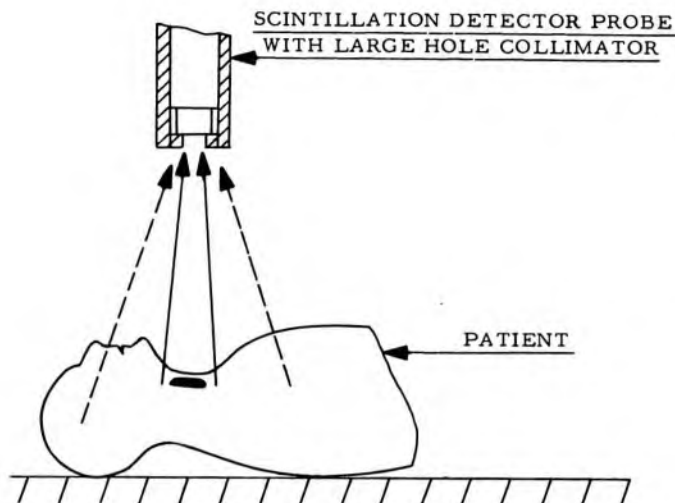
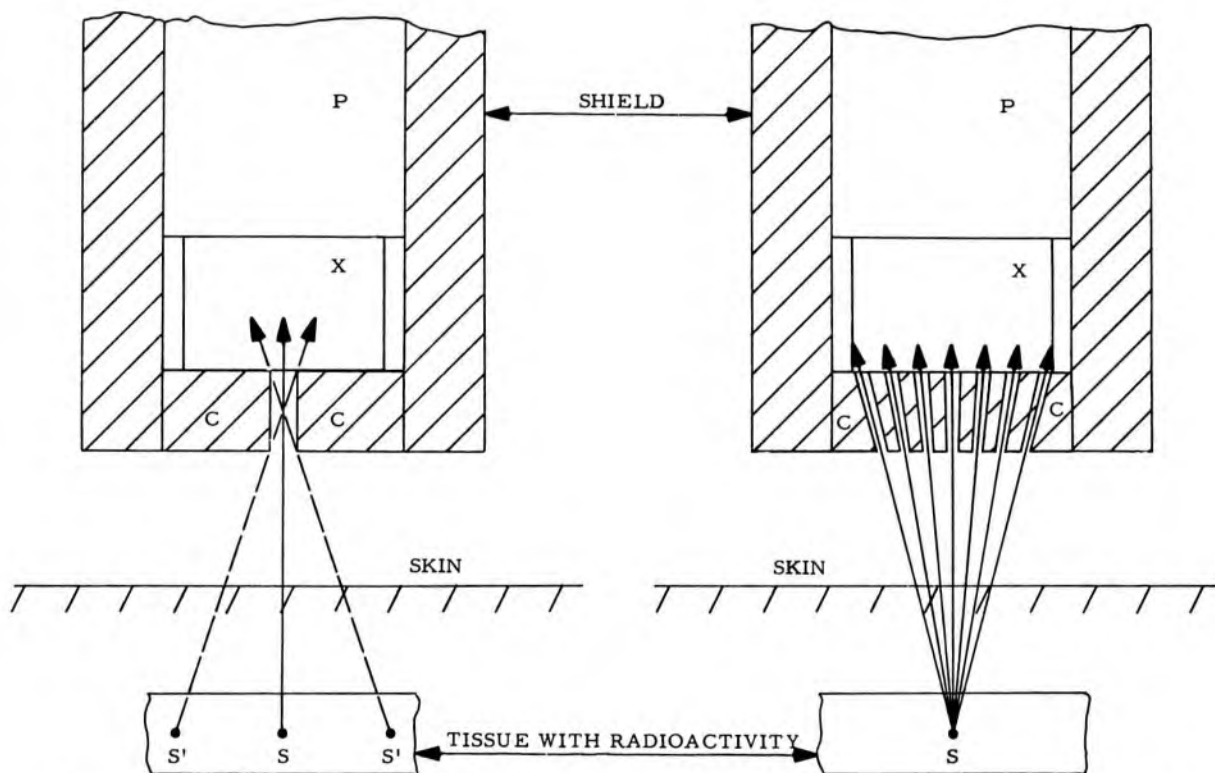


Fig. 1. Some alternative collimation setups. Left. Wide-angle collimation: thyroid uptake measurement. Note large neck area is sampled, but photons from major blood pools of chest and head (dashed lines) are substantially prevented from reaching detector.



Bottom left. Narrow-angle collimation: small hole type. Area of sampling is reduced. However, so is count rate, and penumbra radiation (S') reduces collimation effectiveness. Bottom right. Multi-hole focused collimation: multiple small tapered holes in heavy shielding material. Selectively detects radiation from a local area at a preferred distance, with greater sensitivity than a small hole collimator. Code: C, collimator insert; X, scintillation crystal; P, photomultiplier tube.

NARROW-ANGLE COLLIMATION. In other measurements one wishes to selectively measure count rates from relatively small areas of the patient with minimum count rates from adjacent areas. This technique is used for both fixed probe and mechanical scanning studies. In fixed probe measurements it permits selectively measuring activity from small volumes of tissue, such as large blood vessels, heart chambers, head of the pancreas, etc. In mechanical scanning it is indispensable to obtain the fine resolution needed for evaluating organ structure and function.

Narrow-angle collimation, as it is called, is most simply carried out using a lead cover with a small hole placed over the detector [Fig. 1(bottom left)]. Although this can be made quite effective, it reduces the system sensitivity greatly by reducing the crystal area which receives photons. Some off-center "penumbra radiation" [ray (S')] also reduces the collimator effectiveness. If the collimator thickness is increased to reduce this penumbra, the added weight can be troublesome, and sensitivity may be further reduced by the greater required detector distance.

MULTI-HOLE FOCUSED COLLIMATORS. Multi-hole collimators have been developed to increase the sensitivity of any system for a given degree of collimation [Fig. 1(bottom right)]. They use multiple tapered holes in a block of heavy material, such as lead or tungsten alloy; the axes of the holes intersect at the desired location. Such collimators are now made commercially with 19, 37, and more holes, and some experimental units designed for low photon energy emitters (^{99m}Tc , for instance) have been built with up to 1,045 holes for a 3-inch crystal.

Like grids in x-ray diagnosis, multi-hole collimators are effective only if they absorb off-focus radiation which strikes the hole walls (septa) obliquely. For this reason, such collimators are most effective for low energy gamma emitters, like ^{125}I , ^{99m}Tc , and ^{197}Hg . They are less effective, but still useful, for intermediate energy emitters, like ^{131}I , ^{203}Hg , ^{198}Au , ^{75}Se , and ^{51}Cr . They are of limited usefulness, however, with high energy emitters, like ^{59}Fe , ^{60}Co , and ^{47}Ca .

Focused collimators are effective primarily at one distance; at shorter and longer distances reduced response is observed. Figure 2 shows measured "isoresponse curves" for a focused collimator.¹ The lines are locations of radioactivity yielding the same count rate for a given activity, expressed as percent of the response obtained with material 2.7 inches from the probe and perfectly centered. Precise focal response along the plane AA is obtained, with falloff of response to 10 percent of maximum 5 mm either side of the center. However, if the organ distance is improper, both the sensitivity and discrimination of measurement are reduced. For example, at 3.7" depth, just

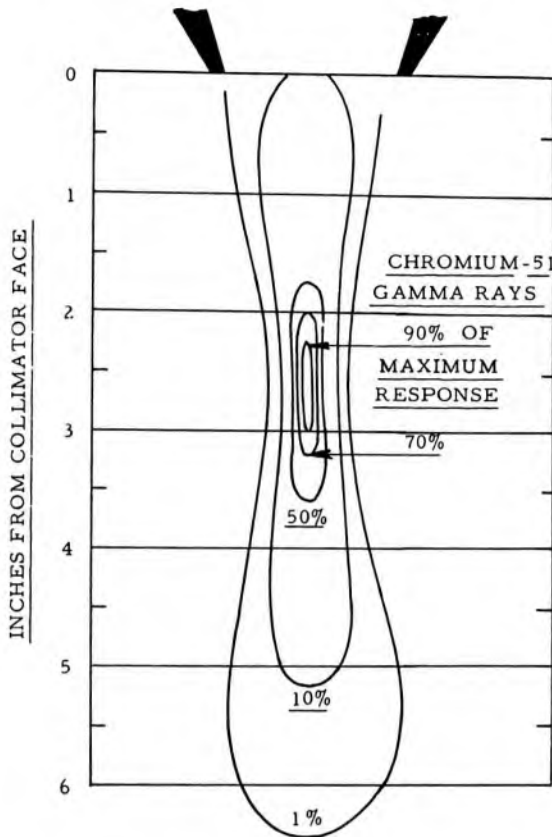


Fig. 2. Isoresponse curves for a multihole focused collimator. The lines show positions below the collimator at top for which 1, 10, 50, 70, and 90 percent of the maximum response is obtained. (Redrawn from Harris et al.¹)

one inch from AA, the central area signal is only 45 percent of what it should be. Moreover, at this distance the response falls off far more slowly: the count rate is reduced to 10 percent of maximum at 15 mm from the center, compared with 5 mm at the proper focal distance. Consequently, contrast is greatly reduced at any off-focus depth.

This criticalness of distance is very important in practice. Not only must the proper distance be selected in positioning the patient, but shifts in position during a measurement or scan become very important. One must also keep in mind the dependence on depth when interpreting studies of organs of finite thickness and variable distance to the detector, such as the liver. Such anatomic variations are substantial and probably set practical limits on the useful collimation precision.

Tissue Attenuation Effects

So far we have not considered the effects of photon attenuation by the patient. In addition to intermediate tissues, the organ itself usually has significant thickness (i.e., liver and brain). Significant attenuation effects occur, particularly with lower energy gamma rays, since much tissue must be traversed on the way to the detector. Consider photons of ^{131}I , $^{99\text{m}}\text{Tc}$, and ^{125}I . These are absorbed about 21, 28,

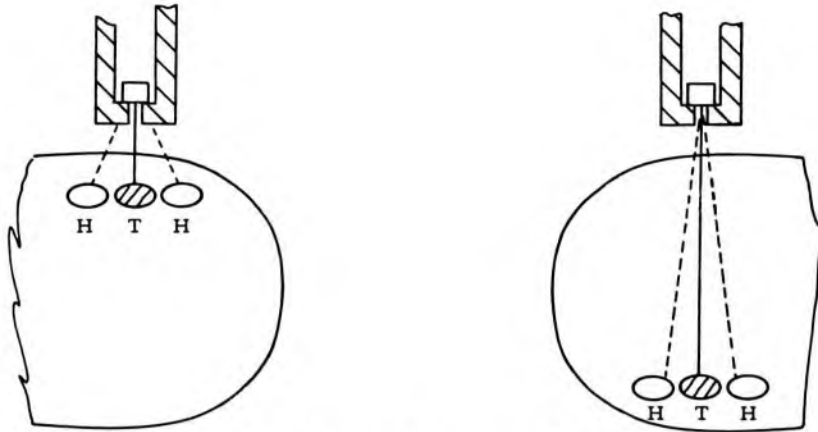


Fig. 3. Scanning is most sensitive when done close to the area of interest. Consider a lateral brain lesion. T represents tumor tissue, H, healthy tissue of different radioisotope concentration. Left. Scanned from proper side:— optimum shielding and distance; H signals attenuated to considerable extent. Right. Scanned from opposite side:—separation of T and H signals is much poorer, for geometric reasons. In addition scatter from both H and T further reduces the image contrast.

and 55 percent, respectively, by 2 cm of soft tissue. Consequently, apparent substantial local differences in tissue concentration may actually be due to differences in organ depth!

Another depth effect is reduction in apparent contrast of the target organ versus adjacent tissue when measurements are made from the wrong side. Consider a brain lesion measured from a proximal (A) versus a distal location (B) in Figure 3. In (A), radiation from healthy tissue H is substantially excluded from the detector; in (B) it is detected because of its more remote location. A scan performed with situation (B) is greatly inferior technically to that of situation (A) in that differences in radioactivity are less apparent. For this reason, many people routinely perform scans from both sides.

What sort of radioactivity concentration differences are needed for detection in a scan? Generally speaking, the activity concentration in a target organ must be much more than twice that in surrounding areas, under even the most favorable circumstances, for unambiguous identification. Of course, many of the factors mentioned above influence the required ratio: organ size, collimation, statistics, depth below the skin, to mention only a few. In general, hot areas are more readily detected than cold ones, so a study seeking greater activity has a better chance of success than one seeking decreased activity, other factors being constant. *It is important to realize that relatively great concentration differences are required to yield barely detectable scan indications.* This aspect cannot be overemphasized in designing scanning techniques with *any* type of scanner.

SCANNING

Very often one is interested in obtaining the distribution of radioactivity in an organ, showing areas of both increased and decreased activity. Scanning provides a useful two-dimensional record of this distribution, within the limitations noted above.

Two basic parts exist to the problem of obtaining a useful scan. The first is most fundamental: radioactive material of proper type must be appropriately administered in a manner which produces a distribution in various organs which is of diagnostic significance. This is the basic problem of radiopharmacology. The second problem is primarily physical: the resulting distribution must be recorded with sufficient fidelity to permit useful interpretations. This is the task of scanning.

In this section we first briefly consider several clinical scanning procedures and their mechanisms of regional concentration. We then discuss various medical scanning systems.

Concentration Mechanisms

Several different types of clinical scanning procedures are listed in Table 1.² There are many ways in which diagnostic radioactivity concentration differences are produced; we shall comment briefly concerning four important ones:

- A. Functional tracer concentration by the organ.
- B. Blood pools, normal and abnormal.
- C. Mechanical trapping of larger tracer particles.
- D. Trapping of tracer inside selectively damaged tumor cells.

SELECTIVE ORGAN CONCENTRATION. Many of the most useful studies involve organ concentrations of tracer material. Thyroid glandular tissue has an affinity for inorganic iodine as well as some ions of similar chemistry, of which $\text{Tc}\bar{\text{O}}_4$ is one of the most important. Although $\bar{\text{I}}$ or I_2 and $\text{Tc}\bar{\text{O}}_4$ are all trapped, however, only iodine is actually converted into thyroid hormone. Hence, $\text{Tc}\bar{\text{O}}_4$ is useful as an indicator of areas of *acquisition* of inorganic iodine but not of *conversion*.

The kidneys concentrate both chlormerodrin and hippuran but excrete only hippuran readily. Both materials are useful for scanning; hippuran is also useful for a combined scan and renogram.

Colloidal materials of very small particle size are useful for localizing reticuloendothelial tissue, which actively removes and fixes it. The liver and spleen general locations can be pinpointed in this

TABLE 1. CLINICAL SCANNING PROCEDURES—
MECHANISMS OF REGIONAL CONCENTRATION²

- A. Functional concentration of tracer by organ
1. Thyroid: trapping of $\text{Tc}\bar{\text{O}}_4$ and $\bar{\text{I}}$ and I_2
 2. Kidney: chlomerodrin and hippuran accumulation; excretion of hippuran
 3. Reticuloendothelial system: liver, colloidal ^{198}Au
 4. Spleen: damaged or aged red cell sequestration (^{51}Cr)
 5. Liver: rose bengal by polygonal cells
 6. Bone: active areas of calcification concentrate Ca^{++} and Sr^{++}
 7. Cartilage: concentrates $\text{S}\bar{\text{O}}_4$ and $\text{Se}\bar{\text{O}}_4$
 8. Parathyroid and pancreas: selenomethionine (^{75}Se)
- B. Blood in pools (human serum albumin, with $\bar{\text{I}}$ or $\text{Tc}\bar{\text{O}}_4$ label)
1. Normal: heart, great vessels, placenta
 2. Abnormal: failure of brain-blood barrier due to tumor, hematoma, infection etc.
- C. Mechanical trapping of larger tracer particles.
1. Microemboli: macroaggregate albumin (MAA) in capillaries, mainly the lung
 2. Aerosols: particles inhaled into alveoli
 3. MAA emboli: peripheral circulation (experimental)
- D. Trapping of tracer in selectively damaged cells:
- Brain scanning following regional perfusion of H_2O_2 solution (see text)

way using ^{198}Au . However, if liver *function* is desired, the site of functioning polygonal cells is of more direct interest, and rose bengal is more useful. In any liver scan, lateral and even posterior views are often needed. In addition, there are evidently wide normal variations in liver contours; these variations must be carefully considered in scan interpretation.

There are several less common scan studies involving organ concentration (Table 1). Areas of bone with active calcium deposition concentrate calcium and strontium radionuclides. Similarly, chondrocytes trap $^{75}\text{Se}\bar{\text{O}}_4$ (which mocks $\text{S}\bar{\text{O}}_4$ chemically), incorporating it in cartilage, thereby identifying areas of hyperactivity in lesions like chondrosarcoma. Selenomethionine similarly fools both the parathyroid and pancreas glands into selectively concentrating it like ordinary methionine, providing a basis for scanning these organs with ^{75}Se -tagged selenomethionine.

BLOOD POOLS. Almost any tracer which is relatively inert to body action can be used in techniques based on local blood accumulation,

the second basic method. Currently, iodine and technetium, as radioactive tags in human serum albumin, are most useful for heart, great vessel, and placental scanning.

In brain tumors the exact tumor concentration mechanism is no doubt more complex, involving other factors than simple hypervascularization, and some preparations concentrate better than others. Failure of the blood-brain barrier is probably involved in the transfer of soluble tracer material to brain tissue.

MECHANICAL TRAPPING OF LARGER TRACER PARTICLES. Suspensions of tracer materials have recently been used for identifying the location of small vessels in which they are trapped. Of course, the material must be readily broken down to restore circulation promptly. This is accomplished by using ^{131}I -tagged albumin in an aggregated form (macroaggregated albumin—MAA), with particles about 20 to 50 microns in diameter. This is administered in appropriate blood vessels and lodges in arterioles and capillaries; it is broken down in a matter of hours without reported injury. The main use for this technique has been in lung circulation scanning. Colloidal tracers (gold, MAA, and technetium in sulphide form) of much finer size (one micron) are used in aerosols to outline alveolar locations. The two studies have been combined to obtain a fuller picture of lung function.

MAA has also been used by Wagner to study peripheral circulation. Once its safety is firmly established, this ingenious technique promises to have many additional uses.

SPECIFIC TUMOR CELL LOCALIZATION AFTER H_2O_2 TREATMENT.³ Malignant cells can apparently be made to specifically concentrate radioisotopes. Finney has shown that malignant cells have defective cellular membranes: they are irreversibly fractured by oxidation.⁴ Injured cells tend to accumulate tracers (through membrane defects) which would ordinarily remain in the extracellular space. Such tracers remain trapped within the cells for many days, long after the extracellular material has been routinely "washed away" and excreted.

In practice, this principle has been found most useful in brain scanning. The brain is regionally perfused with H_2O_2 before addition of the tracer, thereby preparing the tumor cells. The administered tracer is later trapped in these cells, and much remains after two or three days, when most blood-borne radioactivity has been excreted. Since healthy cell membranes are not damaged this way, they do not concentrate the tracer (unless previously damaged from another cause). The result is a scan with minimum tracer in ocular, sinus, muscle, and similar areas which initially mask tumor concentrations because they are rich in blood-borne tracer. Promising preliminary

results have been reported, but the method is relatively new and still under investigation.

Scanning Systems

A bewildering variety of scanning systems confront the prospective user; it is therefore useful to briefly summarize the basic types and discuss their principles and applications.

BASIC SCANNING METHODS. Scanning can be accomplished in several ways. The simplest is to move a collimated probe manually over the patient, taking counts at each of several points and plotting the result by hand ("hand scanning"). This is time-consuming as well as inaccurate and physically tiring since a well-collimated probe is generally heavy and difficult to position and aim accurately.

Mechanical scanning was developed to solve these problems. In addition to a solid mechanical support for the heavy probe, it also provides a motorized system for making successive passes over the area of interest. A recording device is simultaneously positioned over a corresponding location in a parallel plane and appropriate marks made on a paper or film in response to the counting rates. When the probe is over areas of great activity, marks are made in close succession yielding a darker portion of the record; over low activity areas few marks are recorded. Many printing devices have been used, giving an array of dots on paper called a "dot scan." Another type of record is made on x-ray film: each film area is exposed to light illumination proportional to the counting rate over the corresponding sampled area of the patient; the result is a "photoscan."

More recently, new "scintillation camera" devices have been developed. These respond simultaneously to photons from all the areas of interest, and the *total* image is produced constantly during examination of the patient. Cameras have some basic advantages over mechanical scanners but at this writing are not in as widespread use.

Multiple scans can offer more information than single scans, just as multiple versus single roentgenograms. Much research work is in progress along these lines.

MECHANICAL SCANNING. Although the basic setup in mechanical scanning has already been described, some additional points are worth mentioning.

One fundamental limitation of a mechanical scanner is that it samples small areas for only a very short time. There is an inherent sensitivity dilemma: to obtain enough counts from a particular area requires a reasonable sampling time; however, if this is too great (i.e., scanning is too slow) the study is excessively protracted. The

use of a larger scanning crystal (say, 5" versus 3" diameter) helps somewhat. Probably using short-lived pure gamma-emitting materials like ^{99m}Tc is the best solution, as this permits scanning with *millicurie* doses versus 50 to 100 *microcuries*, without excessive patient dosage. An incidental additional benefit is improved collimation since these short-lived materials generally emit lower energy photons. Of course, statistical questions are basic to all radioactivity measurements and arise with camera scanners also.

An annoying problem with mechanical scanners is an occasionally observed shift in the dot pattern along each line, giving a false serrated edge (Fig. 4). This arises from the fact that the detector scans in a reciprocating manner, advancing a step with each reversal. There is a ratemeter and printing lag in registering the dots, so the record of a line is always displaced towards the starting end. Newer machines have design modifications to minimize this effect.

Much variety exists in scan printout schemes. The simple dot

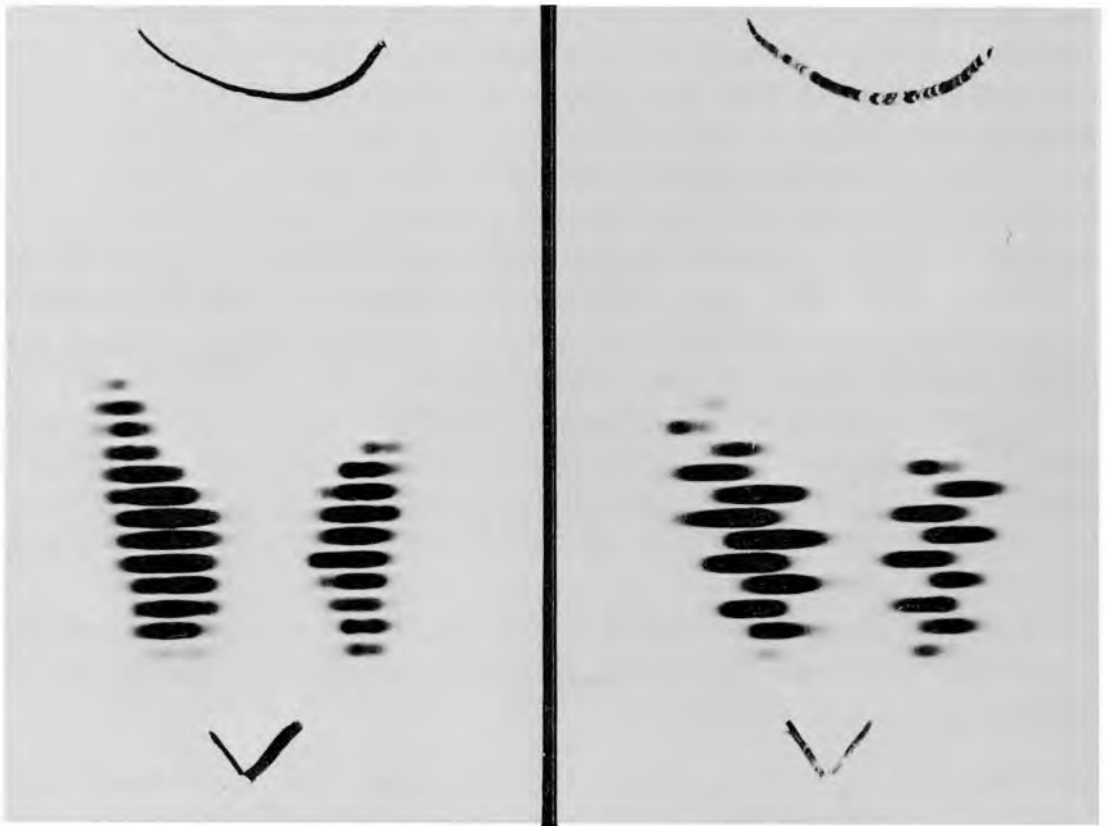


Fig. 4. Serrated edge record of mechanical scanner (Right: extreme example). Note the edges of the thyroid image are ragged because the scanning motion is reciprocating and scanning factors poorly chosen. This effect is most noticeable with very rapid scan speeds and slowly responding ratemeters. It should be stressed this is an extreme example and useful results can generally be obtained with proper use of such scanners.

scans contain the desired information, but users most often prefer somewhat larger dots. Photoscanning was developed to make the picture look more like a roentgenograph, with more familiar shades of gray.⁵ The search for greater precision of boundary identification has led to the development of special printing systems in which different colors are used for different ranges of measured activity. The question appears to be one of subjective preference, since the eye can differentiate at least as many shades of gray as colors economically available. Perhaps a more significant development is the use of a better printout system than film to obtain a more linear gray scale. It will be recalled that the density versus exposure curve for film activated by light is nonlinear. Kuhl has suggested that a more linear printout system than film can improve the accuracy of display.

CAMERAS. A mechanical scan is inherently inaccurate because the initial part of the record is obtained at a different time from the last part. In a camera scan all parts of the image are recorded simultaneously, a fact which can be important in measuring rapidly changing distributions, and when short-lived radioisotopes are used.

Four types of cameras have been recently developed:

1. "Scintillation camera" (Anger)
2. "Autofluoroscope" (Bender and Blau)
3. "Image tube scintillation camera" (Ter-Pogossian)
4. "Spintharicon camera" (Horwitz)

All of these units are relatively new at this writing and must be considered ingenious devices not yet fully evaluated. However, the field is developing rapidly, so a few brief comments regarding the basic operating principles and more obvious potentialities and limitations of these units are in order.

The first two units employ NaI(Tl) crystals and photomultiplier tubes; the other two use fluoroscopic screen and gas detectors. The latter are therefore optimally suited to measure low energy gamma emitters like ¹²⁵I, ^{99m}Tc, and ¹⁹⁷Hg. This limits their greatest usefulness to relatively superficial structures because of greater tissue attenuation of low energy photons. This deficiency can be overcome to some extent by scanning from both sides and laterally, and it is probable that imaginative application will extend their useful range.

All these "cameras" provide a final image on a photographic film, usually employing a Polaroid camera.

ANGER CAMERA.⁶ The Anger camera employs a single large 1/2" thick NaI(Tl) crystal, 11 1/2" in diameter (Fig. 5). Light flashes are viewed by an hexagonal array of 19 photomultiplier tubes. These tubes are placed a slight distance away from the crystal and view overlapping areas. The distribution of pulse sizes from the photomultiplier

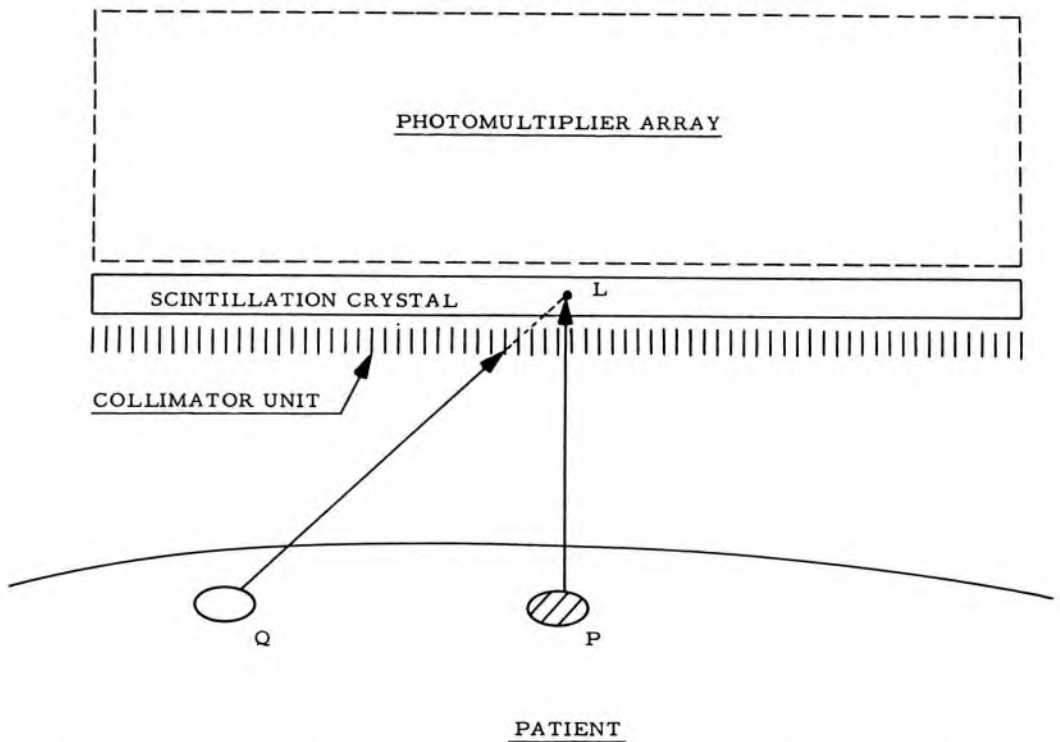


Fig. 5. Collimation is critical with the Anger camera. Note that light can be produced at L in the flat scintillation crystal almost as well by undesired rays from Q as by those from P. To prevent this an effective collimator is needed below the crystal. (Redrawn from Anger and Bender.⁶)

tubes corresponds to a particular light pulse location in the crystal and hence to the location in the patient where the initial decay occurred. The location of each flash is computed from the photomultiplier pulse sizes and this information communicated to a cathode ray oscilloscope. By suitable electronic methods a brightness distribution on the cathode ray oscilloscope tube face is built up corresponding to the radioactivity distribution.

Only the photopeak signals are used, to minimize scatter and other unwanted signals. About 1,000 to 100,000 dots are recorded, corresponding to a reasonable period of observation (a few seconds to a few minutes, depending on the radioactivity). Collimation is, of course, crucial since the detecting system itself cannot differentiate between an oblique and a direct ray absorbed at the same location in the crystal (see Figure 5). Several collimators have been designed for this unit: two "pinhole types" for thyroid work and two multihole units for low and higher keV gamma emitters.

Anger has also constructed a special version for positron emitter scanning. It employs *two* detector units with a coincidence circuit arrangement, so only the 0.511 MeV annihilation rays are recorded. This results in a very effective electronic collimation. Although excel-

lent scans are obtained with these special Anger cameras they are costly and complex in construction, use, and maintenance, currently limiting their use to a few larger centers.

The Anger camera is limited by the necessity for a $\frac{1}{2}$ " maximum crystal thickness. If a thicker crystal is used, resolution suffers; in addition, there is loss of information near the edge of the field.

AUTOFLUOROSCOPE.⁷ The Bender-Blau "autofluoroscope" minimizes crystal resolution problems by employing a rectangular mosaic of 260 NaI(Tl) crystals; each $\frac{3}{8}$ " in diameter and 2" long. They are packed in an array $6'' \times 9''$ in size (13 and 20 crystals on a side, with centers 1 cm apart). These crystals are efficient absorbers, and the array lends itself readily to efficient plastic "light-pipe" communication of light flashes to the photomultipliers. Figure 6 shows the collimator (A), crystal (B), and light-pipe (C) parts of the system. An arrangement is employed to permit identifying which row and file crystal has been energized, by use of light pipes conveying light to an array of 33 photomultiplier tubes. The data are ultimately processed to display the scan on a large cathode ray tube. The system resolution is very

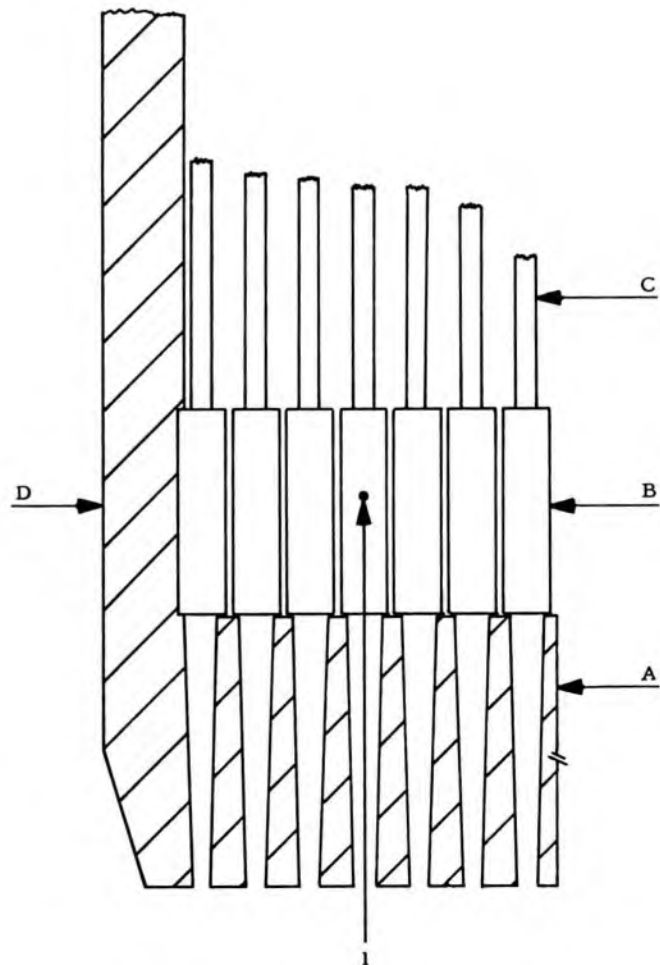


Fig. 6. Bender-Blau Autofluoroscope—Detector System. Shown are the collimator (A) with tapered holes, the cylindrical NaI (Tl) detector crystals (B), with light pipes behind the crystals (C) and lead shield (D). Desired gamma rays (1) are almost all absorbed by the 2-inch thick crystals, permitting use of high energy gamma emitters. Electronic correlating and coincidence circuits combine multiple pulses to infer locations of the source of the rays. (Redrawn from Anger, H. O. and Bender, M. A.⁷)

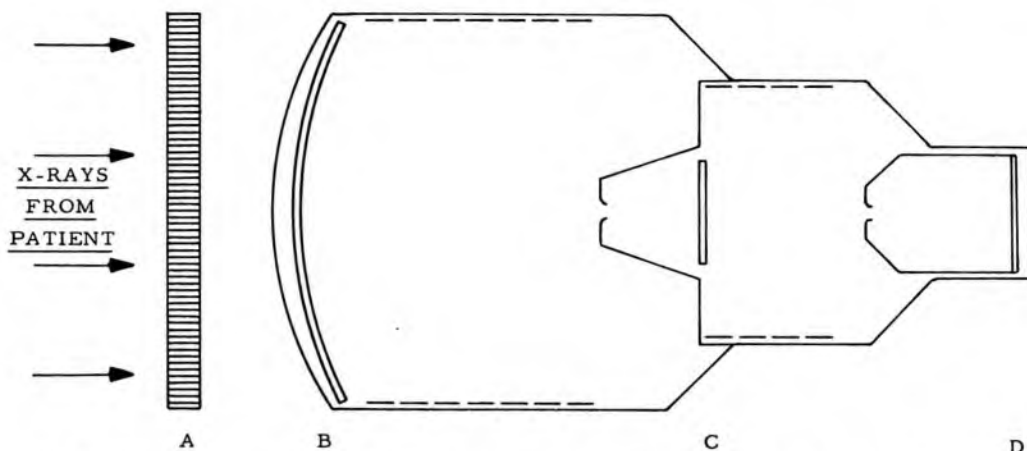


Fig. 7. Image tube scintillation camera (see text). (Redrawn from Ter-Pogossian et al.⁸)

good, limited primarily by the fluorescent crystal diameter. Its good collimation and response to high energy photons are also inherent advantages.

IMAGE TUBE SCINTILLATION CAMERA.⁸ Both of the crystal camera systems use quite complex detector systems and associated electronics. When low energy gamma emitters are useful, the "image tube scintillation camera" of Ter-Pogossian is perhaps somewhat closer to meeting practical requirements than the crystal cameras.

The operation of the system is attractively simple. (Figure 7 shows a schematic representation.) The multi-hole collimator A limits rays reaching the fluorescent screen at B to direct ones, in order to permit proper imaging of the radioactive distribution. Screen B is similar to that in the ordinary x-ray image intensifier tube. Its fluorescent coating on the left side of the thin glass mount glows in response to the incident gamma rays; the resulting light in turn liberates electrons from the photoemissive surface on the other side, yielding an electron image of the radioactivity distribution.

From here on, the system is simply a two-step intensifier. The image on C is an intensified version of that at B. As in ordinary intensifier tubes, this has been accomplished by both electron acceleration and minification. The visual image at C is coupled by fiber optics to a second photoelectric surface, which is the cathode of a purely *optical* image intensifier. (For simplicity the details of focusing electrodes and electric connections are omitted.)

The net result is an over-all brightness gain on phosphor (D) of 50,000 times. This great a gain is necessary here because signal levels are so small. But what of statistical noise? Unlike the x-ray situation, scanning times of up to minutes are possible. The use of high activities of tracers also greatly helps the statistics. In addition, alternative

scanning systems yield images of relatively poor resolution, so the requirements are less demanding than those of x-ray images.

SPINTHARICON CAMERA.⁹ The "spintharicon camera" is a gas discharge system which has inherently excellent resolution. It makes use of the fact that a gas discharge can produce considerable light. In Figure 8 the filling gas is voltage stressed in a manner similar to that in a G-M tube. When a gamma ray enters and interacts with the silver coating on the glass, electrons ejected from the silver trigger a spark discharge. This spark is extremely bright, and its light is readily recorded by a Polaroid film even when viewed end-on. Argon-neon mixtures are used, with alcohol and iodine vapors as quenching agents.

The sensitivity of this instrument is limited by its low detector efficiency (below 1 percent). It is likely that its sensitivity lies between that of mechanical systems and the other cameras. Its potential resolution is excellent, however, particularly with low photon energies which permit good collimation.

The spintharicon camera has several attractive features. It is basically relatively simple and inexpensive; its potential resolution is quite good; and finally, it is a light, compact device which can be readily placed close to the patient's skin, helping to some extent to compensate for its inherent relative insensitivity.

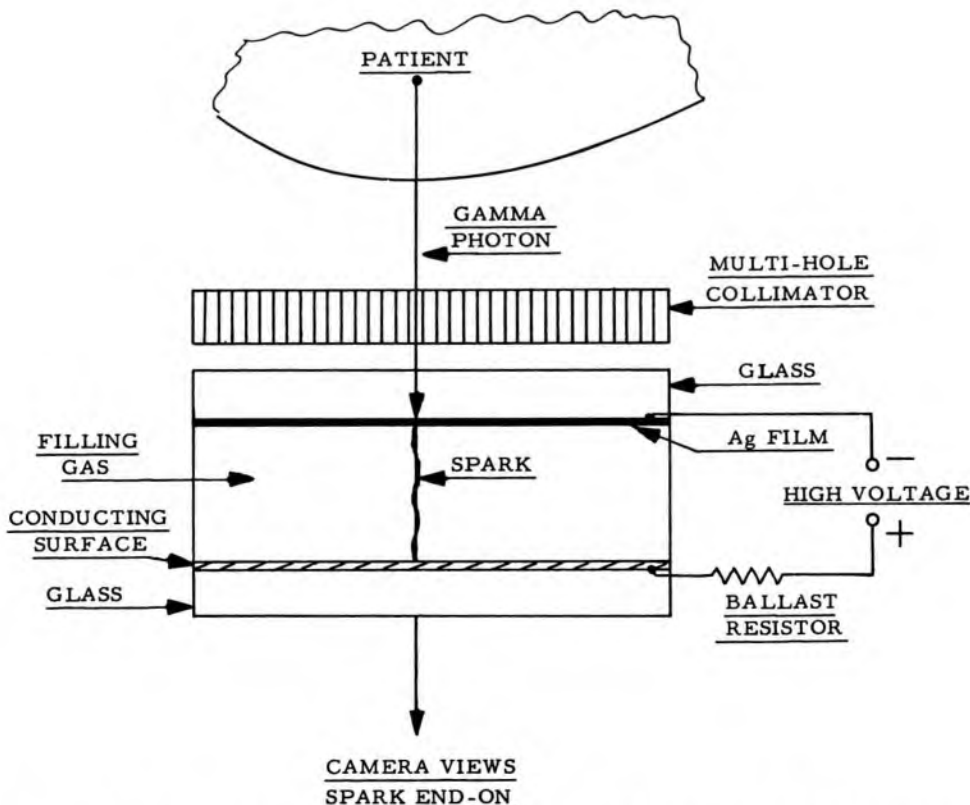


Fig. 8. Spintharicon camera (see text). (Redrawn from Horwitz et al.⁹)

MULTIPLE SCAN STUDIES. Economic considerations normally limit the physician to one scan at a time. However, just as multiple views and body section studies are often invaluable in radiography, they potentially have their place in radioactivity scanning studies as well.

Two detectors are used in positron scanning to obtain a single record, as previously described. The excellent "electronic collimation" of such systems can potentially greatly increase scanning resolution.

Scans obtained with focusing collimators usually suffer from the fact that maximum response is limited to a rather thin section of the patient. This can result in loss of details on both sides of this slice of tissue. The problem can be greatly benefited by using twin detectors, one on each side of the patient, which move together. (The photomultiplier pulse signals are added together.) If the collimators and crystal separations are appropriate, a relatively constant response is observed with depth for emitters of intermediate energy like ^{131}I .¹⁰

An even more elaborate system using *four* detectors has been developed. It simultaneously scans over both sides and front and back of the patient, yielding four records. Although such a unit is, of course, much more costly, one cannot deny the great benefit in a busy department of obtaining maximum information in a reasonable time period.

All of the above systems yield either a better single scan or multiple views. Kuhl has extended scanning to body section studies, accomplishing this in a manner analogous to x-ray tomography by moving the scanner about the patient in various ways.¹¹ (Hisada reports an interesting two-probe unit which yields section-like results with less elaborate apparatus.¹²) Quite impressive and unambiguous results have been obtained in liver and brain scans, and there is no doubt that scanning tomography has much to offer in increased yield of diagnostic information. The expensive and complex equipment required at present, however, limits such studies to a few institutions.

IN-VIVO BETA AND WHOLE BODY COUNTING

Scintillation probe and scan measurements include the most frequent clinical tracer studies. In-vivo beta counting, however, has yielded interesting diagnostic results in some people's hands, and whole body counting has other important applications.

In-vivo Beta Measurements

By this term we have in mind measuring radioactivity *in vivo* with detectors placed directly alongside tissue to be studied. Much

work has been carried out on the localization of tumors, using ^{32}P as phosphate ion. Intravenously administered ^{32}P is selectively concentrated by some tumors, as well as by benign tissue under unusual circumstances such as inflammation. Special beta-ray detectors have been employed. For eye tumors tiny flat G-M counters are useful, particularly behind the eye. Robinson has developed a very fine diameter needle probe for detecting ^{32}P rays during brain surgery, to localize tumor tissue. Finally, gastric tumor localization has been attempted with ^{32}P using scintillation crystals at the end of glass fiber light pipes, p-n junction diode detectors, and inflatable balloons with photographic emulsions on the inside.

Recently a new p-i-n silicon junction diode detector has been reported. This unit is 4 mm in diameter and 6 mm long. Presumably the time is not far off when even smaller units will be developed to fit into catheters, permitting in-vivo detection of radioactive isotopes in blood vessels of many body organs. The relative insensitivity of these detectors, however, currently limits their usefulness to situations in which patient radiation dosage is not of primary concern.

Whole Body Counting

There are occasions when it is desirable to know the total amount of radioactivity due to a particular radionuclide in the body. For example, the amount of ^{226}Ra , ^{137}Cs , or ^{131}I in a person eating certain foodstuffs over a given time may be of great interest. The body retention of certain elements may also be an index of physiologic activity (for example, ^{47}Ca retention in evaluating bone metabolism, ^{59}Fe retention in anemia, etc.). Instruments have been developed for such measurements, and we shall briefly describe two of them.¹³

Regardless of how the measurement is carried out, shielding is very critical in whole body counting because extremely small amounts of radioactivity must be detected, and counting rates may be comparable with or even smaller than background radiation levels. The problem is complicated by the fact that natural lead has considerable radioactive contamination (since its natural isotopes are end products of radioactive decay). For this reason, steel shielding is usually employed (usually old battleship steel). Five or six inch thick shields are required to permit scintillation measurement of ^{40}K and fallout ^{137}Cs in patients. Measurement rooms usually contain 25 to 100 tons of steel and are therefore very expensive although recently compact setups requiring more reasonable amounts of shielding have been reported. Instruments for whole body measurements of small animals can be made much smaller, and commercial units are available for this purpose.

Human whole body counting has been carried out using both crystal and liquid scintillation detectors. The Los Alamos liquid scintillation unit centers the patient along the axis of a hollow cylinder which contains 140 gallons of a toluene solution of terphenyl and POPOP liquid scintillators. One hundred and three 2-inch photomultipliers are arrayed uniformly around this annular solution volume to detect the scintillations. The entire assembly is surrounded by a 20-ton processed lead shield 5" thick. Although the spectral resolution is not as good as that of crystals, the liquid detector offers extremely efficient geometry and is very sensitive.

The solid crystal whole body unit (Argonne National Laboratory) uses a single, very large NaI(Tl) crystal (8" diameter \times 4" thick). This crystal is placed about 40 cm above the subject in a geometry to assure good sampling of the major anterior portions of the body: head, trunk, and thighs. Multiple photomultiplier tubes on the large crystal feed pulses to a multi-channel analyzer.

Why are there two types? The Los Alamos unit (liquid scintillator) is quite sensitive and adequately separates ^{40}K and ^{137}Cs , the most important long-lived radionuclides in the body. The Argonne unit (solid crystal), although less sensitive, has better spectral resolution. It can therefore specifically identify many other radionuclides present.

Units for small animal work are much cheaper but essentially based on the same principles as the larger ones. One commercial unit employs liquid detection; another uses two moderate-sized crystals between which the animal is placed during counting. In addition to small animal work, they have been used for arm radioactivity measurement. Both units have extensive shielding; in fact, they are so heavy they can be moved about only slowly and with considerable effort.

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Clinical Radionuclide Dosimetry

The radiotherapist has a wide range of radiation sources available from which to choose (Table 1). From a purely practical point of view, x-ray therapy has two basic advantages over radioactivity therapy. It is far more versatile, offering treatment distances from 1.8 to over 100 cm, and effective beam energies of a few keV to over 10 MeV, with $d_{1/2}$ values of a few millimeters to over 15 cm. In addition, x-rays are more easily controlled: they can be relatively easily collimated and turned off and present no radioactive contamination hazard.

Nevertheless, radioactive materials make their own very important contribution to radiotherapy. Actually, most supervoltage quality external beam irradiation is carried out using cobalt-60 machines; there are more than 1,000 such teletherapy installations throughout the world. Radium has been used for direct application to tumors almost since its discovery. Sources have generally been used in and alongside tumor tissue, with all but gamma rays screened out. Such use is called brachytherapy, to differentiate the procedure from other applications. (*Brachy* is Greek for *short*, and the term literally means short-distance therapy.) Since 1946, other man-made radioactive materials, both solid and in solution, have also been used in addition to radium.

As in the use of x-rays, human application of radioactive material requires careful dosage evaluation. This chapter discusses the principles of dosimetry in the therapeutic application of radionuclides. Our coverage is in three sections:

1. Radium and other brachytherapy sources
2. Brachytherapy dosimetry principles
3. Other radionuclide dosimetry.

TABLE 1. RADIATION THERAPY MODALITIES

A. X-ray and Teletherapy Sources

1. Superficial therapy	Below 60 pkV: 1.8 to 30 cm SSD, very low HVL. 80–130 pkV: about 20 cm SSD, 1 to 4 mm Al HVL.
2. Intermediate therapy	140 pkV to 170 pkV: about 50 cm SSD, 0.5 to 1.0 mm Cu HVL.
3. Harder orthovoltage	180 pkV to 500 pkV: about 50 cm SSD, 2 to 5.0 mm Cu HVL.
4. Supervoltage (longer SSD, normally)	Cesium-137 teletherapy: 5.4 mm Pb HVL. *2 MV x-rays: 7.5 mm Pb HVL. Cobalt-60 teletherapy: 10 mm Pb HVL. *4 to 8 MV x-rays: (linear accelerator) 11.3 to 13.3 mm Pb HVL. *20 MV x-rays: (betatron) *31 MV x-rays: (betatron)

B. Localized Radionuclide Sources

1. Brachytherapy	Radium. Radium substitutes: ^{60}Co and ^{137}Cs . Short-lived sources: ^{222}Rn , ^{192}Ir , and ^{198}Au .
2. True solutions	^{131}I and ^{32}P in true solutions.
3. Colloidal solutions and suspensions	^{198}Au and $\text{Cr}^{32}\text{PO}_4$.
4. Beta applicator	^{90}Sr eye applicator.

*Electron beams produced by these machines have also been used. In addition, some experimental work has been done with proton and neutron beams produced using high-energy charged particle generators. (See Chap. 16.)

RADIUM AND OTHER BRACHYTHERAPY SOURCES

Recall that the basic objective of radiation therapy is to produce maximum tumor injury while minimizing that to nearby healthy tissue. Theoretically, the best method of accomplishing this is to place the radiation source in or at least alongside the tumor tissue, so divergence attenuation protects surrounding structures. Since x-rays originate in x-ray tubes of moderate size, the source is generally outside the body, and deep tumors are necessarily treated by multiple portal or rotational techniques. Superficial tumors may be treated with beams of low HVL values and at short or moderate distances to protect underlying structures. From the nature of x-ray production, however, x-ray sources are inevitably significant distances from tumors being treated.

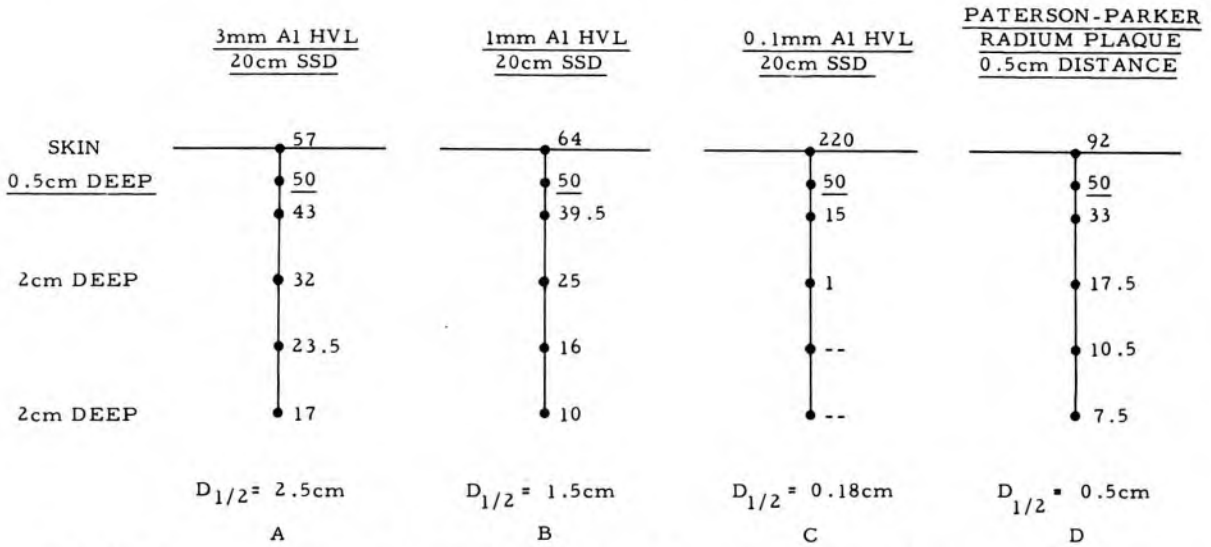


Fig. 1. Comparison of depth dose data: four superficial therapy methods, all with 10 cm² treated area. The very soft x-rays of modality (C) have the most rapid falloff with depth. However, they are suitable only for very superficial lesions. For thicker lesions, radium (D) is the best over-all treatment method from point of view of depth dose. For 50 r delivered to a 0.5 cm depth, falloff beyond this depth is greater than in A and B, sparing deeper structures (see text).

Radium and other radioactive sources are therefore better suited to treatment of superficial and surgically accessible deeper lesions. In superficial therapy of larger lesions radium applicators (see below) provide excellent dosage distribution (Fig. 1). An additional advantage of such "radium molds" over x-ray beams exists when bone or cartilage lies near the skin because the \bar{f} value of radium rays is much lower than that of superficial therapy x-rays. Surgically accessible lesions are treated by direct insertion of sources in tumor tissue and by special molds and other applicators which accurately position sources alongside tumors in body cavities. We shall have more to say concerning these procedures.

Unquestionably, the best radiotherapy involves considerable use of brachytherapy, particularly for certain common cancer sites. Practical difficulties, however, limit its most economical application to larger institutions where a sufficiently great case load makes practical the maintaining of suitable sources, facilities, and personnel. Considerable technician and physician time is required for good work; surgical procedures are required for all but superficial lesions; hospitalization is normally necessary, and radiation hazards are not insignificant. Of all brachytherapy sources, those containing radium are thus far the most frequently employed because radium's long half-life (1,622 years) makes it both convenient and economical to use. In addition, there is a great body of clinical experience with radium. As we shall

see below, however, radium contributes its own unique safety problems.

In this section we shall discuss the origin of radium gamma rays, practical aspects of radium source use, and other radionuclide brachytherapy sources.

Origin of Gamma Rays from Radium Sources

Radium itself is a virtually pure alpha emitter (p. 407), and only 1.2 percent of its disintegrations yields a low energy gamma photon of 180 keV. Yet a radium source produces useful gamma rays ranging in energy from 241 keV to 2.198 MeV. Since these gamma rays do not come directly from the disintegration of the radium atoms, it is evident the *product* atoms must produce them. Actually, radium is the sixth member of a rather long series of radioactive elements originating with uranium-238 and ending with stable lead-206. The most useful gamma rays come from the ninth and tenth members of the 15-element series.

We therefore shall discuss this series and consider the concept of "radioactive equilibrium" which profoundly affects radium dosimetry.

THE URANIUM-238 SERIES. Figure 2 shows the complex series of radioactive transformations by which ^{238}U produces radium and ultimately terminates with stable lead-206. In each transition the disintegrating radionuclide is called the "parent," the new atom the "daughter." ^{238}U is evidently the grand matriarch of this series. In the figure an arrow pointing *down* represents an alpha, and an arrow pointing *right*, a beta disintegration. Note some transitions also involve gamma emission (indicated by γ); of these, all produce infrequent and/or low energy gamma rays, except for ^{214}Pb , ^{214}Bi , and ^{234}Pa . Further information on radium series elements is given in Tables 2 and 3.¹

In many radioactive transformations the new atom is stable. For example, ^{32}P becomes ^{32}S ; ^{198}Au , ^{198}Hg ; and ^{131}I , ^{131}Xe , all of which products are stable. In most other cases one or two radioactive steps may occur, but generally a relatively simple series is involved. However, the three natural heavy radioactive elements (^{238}U , ^{232}Th , and ^{235}U) all produce rather long radioactive series apparently because these very heavy atoms have grossly too many neutrons to be stable with the number of protons they contain. For example, ^{206}Pb is stable, with 82 protons and $206 - 82 = 124$ neutrons. But ^{238}U , with only ten more *protons*, has 22 more *neutrons*. The resulting search for stability involves a long series of particle ejections before nuclear peace is achieved.

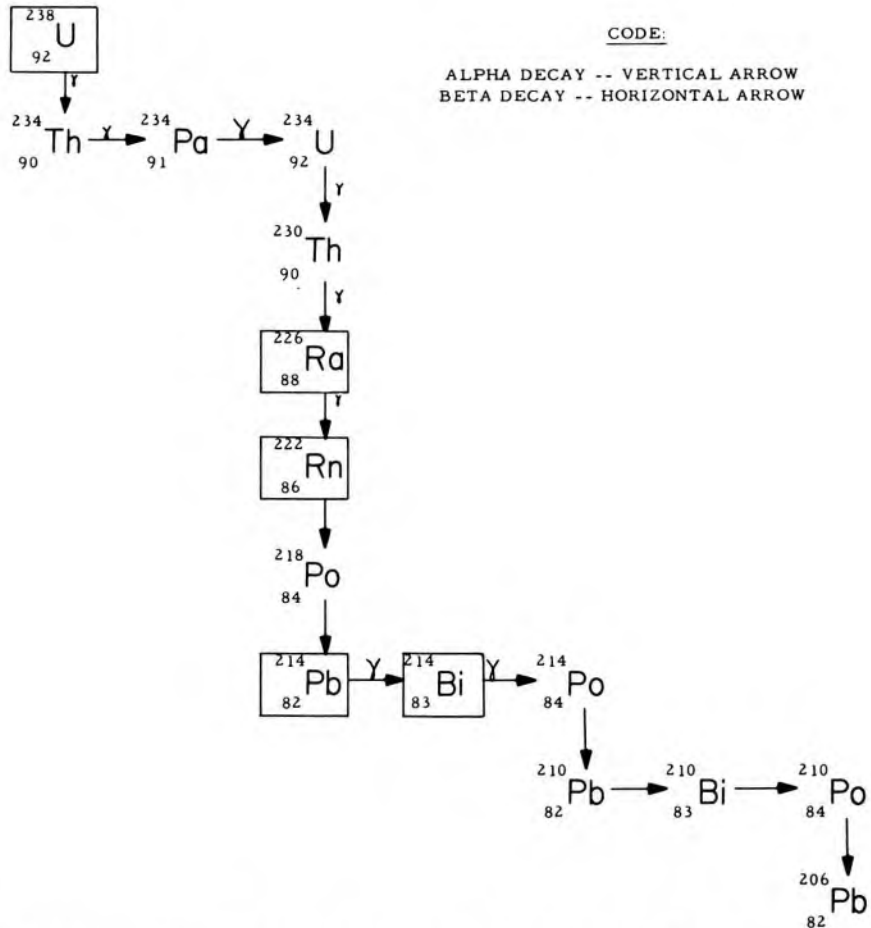


Fig. 2. The uranium-238 radioactive decay series. There are 15 nuclides involved. Alpha disintegrations are represented by vertical, beta by horizontal arrows. Gamma rays are shown by the symbol γ . Most are very weak; three are strong and are indicated by larger γ symbols. The five radionuclides of most interest in this book are shown in boxes.

Only five of the 15 nuclides in Figure 2 are of direct interest to us here; they are itemized below with an indication of their relevance to clinical use.

^{238}U —The source of radium.

^{226}Ra —Sealed in gold or platinum alloy containers as RaSO_4 salt, this nuclide is the source of the radionuclides which emit useful gamma rays.

^{222}Rn —An indispensable intermediate in the production of the desired gamma emitters. Since radon is a gas, it can be lost by diffusion from poorly sealed containers.

^{214}Pb —(RaB). One of the useful sources of gamma rays. Eighty-two percent of its disintegrations produce gamma rays. These are of intermediate energy (Table 2) and useful in radiotherapy.

TABLE 2. NUCLIDES IN THE RADIUM SERIES¹

Nuclide		Half-Life	Energy of Rays—MeV		
Old Jargon	Preferred Symbol		Alpha	Beta	Gamma
Ra	²²⁶ Ra	1,622 yr	4.79–98.8% 4.61–1.2%	—	0.18–1.2%
Rn	²²² Rn	3.83 day	5.49	—	—
RaA	²¹⁸ Po	3.05 min	6.00	—	—
RaB	²¹⁴ Pb	26.8 min	—	0.65	.241–11.5% .350–45% .294–25.8%
RaC	²¹⁴ Bi	19.7 min	—	3.17	.607–2.198 (See Table 3.)
RaC'	²¹⁴ Po	164 μsec	7.68	—	—
RaD	²¹⁰ Pb	19.4 yr	—	0.017	—
RaE	²¹⁰ Bi	5.0 day	—	1.17	—
RaF	²¹⁰ Po	138.4 day	5.30	—	—
RaG	²⁰⁶ Pb	STABLE	STABLE	STABLE	STABLE

TABLE 3. THE MAIN GAMMA RAYS OF A SEALED RADIUM SOURCE¹

Disintegrating Nuclide	Gamma-Ray Energy MeV	Percent of Disintegrations Releasing this Photon
²²⁶ Ra	0.18	1.2
²¹⁴ Pb (RaB)	0.241	11.5
	0.294	25.8
	0.350	45.0
	0.607	65.8
²¹⁴ Bi (RaC)	0.766	6.5
	0.933	6.7
	1.120	20.6
	1.238	6.3
	1.379	6.4
	1.761	25.8
	2.198	7.4

²¹⁴Bi—(RaC). Produced by decay of ²¹⁴Pb. Releases 8 different gamma rays, some of great energy and hence penetration (Table 3). These are also useful in radiotherapy.

More than one gamma ray is emitted in some transformations. (Recall Figure 2 in Chapter 9.) On the *average*, ²¹⁴Bi produces 1.455 useful gamma rays per disintegration, for example.

RADIOACTIVE EQUILIBRIUM. Consider what happens when 10 mg of pure radium salt is sealed in a capsule. At the start there are no daughter elements because there has not been time to accumulate them. Radium decay starts immediately, however, producing ^{222}Rn by decay of ^{226}Ra atoms. From the definition of the decay constant (Chap. 9), we may write:

$$R_0 = \lambda_0 N_0 = \frac{0.693 N_0}{T_0} \quad (12-1)$$

where R_0 = rate of decay of radium atoms
 λ_0 = radium decay constant
 N_0 = number of radium atoms present
 T_0 = radium half-life.

T_0 is very great (1622 yrs.). For example, it takes 25 years for 1 percent of the radium in a source to decay. Hence, radon is produced at a substantially constant rate during, say, a therapist's lifetime.

As indicated above, radon is a gas. Were it stable it would accumulate and perhaps build up great pressure in the container. However, radon is also radioactive, with a relatively short half-life. The rate of radon decay is governed by *its* decay constant, so one may write:

$$R_1 = \lambda_1 N_1 = \frac{0.693 N_1}{T_1} \quad (12-2)$$

Here subscript (1) refers to radon, the first daughter product of radium. Initially $N_1 = 0$, so the radon decay rate is zero. (Obviously, radon atoms cannot decay before they exist.) As time goes by, N_1 increases due to the decay of radium atoms, so R_1 increases correspondingly. It reaches a maximum value when the number of radon atoms is just that for which the radon decay rate exactly equals its production rate. That is, $\lambda_0 N_0 = \lambda_1 N_1$. At this time there is "equilibrium" between production and decay of radon. R_1 cannot exceed R_0 but only approaches it as a limit.

More generally, the same relationship exists among subsequent daughters if we wait long enough for them to achieve their maximum activity:

$$\lambda_0 N_0 = \lambda_1 N_1 = \lambda_2 N_2 = \lambda_3 N_3 \text{ etc.} \quad (12-3)$$

Since $\lambda = \frac{0.693}{T}$,

$$\frac{N_0}{T_0} = \frac{N_1}{T_1} = \frac{N_2}{T_2} = \frac{N_3}{T_3} \text{ etc.} \quad (12-3')$$

This says, in effect, that the equilibrium number of atoms of a nuclide in the series is proportional to its half-life.

Note that in (12-3) each (λN) term is the decay rate of the nuclide involved. *Radioactive equilibrium may hence be defined as a situation in a radioactive series in which radioactive parents and daughters decay at the same rate (i.e., have the same number of millicuries or microcuries).* For the 10 mg of radium situation, there are 10 millicuries of activity of ^{222}Rn , ^{214}Pb , and ^{214}Bi at equilibrium.

TIME TO ACHIEVE EQUILIBRIUM. Since the initial activity of any particular daughter is zero, it is of interest to ask how long must one wait for equilibrium to be achieved for a particular radionuclide. In general, this time depends on the half-life of the element being produced. For example, after a radium source is sealed, the radon achieves half its equilibrium strength in 3.83 days, $\frac{3}{4}$ in $2 \times 3.83 = 7.66$ days, $\frac{7}{8}$ in $3 \times 3.83 = 11.49$ days, etc. In about 30 days over 99 percent of the activity is reached. In general, 99 percent of any daughter is accumulated in about 7 of its half-lives. Since the half-lives of ^{218}Po , ^{214}Pb , and ^{214}Bi are all relatively short, they too will then have reached 99 percent of their maximum activity within 30 days. Thus, a radium source is normally calibrated at least one month after sealing.

It follows from Figure 2 and (12-3) that at equilibrium a radium preparation can produce a total of 5 alpha, 4 beta, and several gamma rays for each radium disintegration! This is because then not one, but nine atoms may disintegrate! The *total mass* of the preparation must remain substantially the same, however, since it is sealed and none can be lost (Actually, a tiny amount of the original radium mass is transformed into *energy*.) The *total number* of the daughter atoms at any time is identical with the number of radium atoms that have decayed since the latter were simply transformed into the former. The great activity is possible only because λ values are very great relative to that of radium.

PRACTICAL ASPECTS. Three interesting questions are often asked in dealing with the radioactive equilibrium of the uranium-238 series.

First, ^{238}U is often used as a shield in teletherapy machines because its great density and Z values make it an effective absorber of gamma rays. Since ^{238}U is itself radioactive and the ultimate source of radium, one might ask how safe a procedure this is.

^{238}U is a pure alpha emitter (Fig. 2), so only its daughters (particularly ^{234}Pa , ^{214}Pb , and ^{214}Bi) yield significant gamma emission. The enormous half-life of ^{238}U (248,000 years) limits the accumulation of its descendents, like radium-226, to very tiny amounts in, say, 25 or 50 years (roughly $1/5,000$ of the activity of the original ^{238}U). The ^{234}Pa , however, rapidly acquires the ^{238}U activity because its half-life

TABLE 4. REQUIRED MASS TO YIELD 1 μ Ci AND RELATED DATA FOR SEVERAL RADIONUCLIDES

A. Uranium-238 Series			
Nuclides	Half-Life	Average Life	Micrograms per Microcurie
^{238}U	4.5 billion yr	6.5 billion yr	2.92×10^6
^{226}Ra	1,622 yr	2,340 yr	1.000
^{222}Rn	3.83 d	5.53 d	6.36×10^{-6}
^{214}Pb (RaB)	26.8 min	38.7 min	0.0297×10^{-6}
^{214}Bi (RaC)	19.7 min	28.4 min	0.0218×10^{-6}
B. Other Radionuclides			
^3H	12.3 yr	17.75 yr	0.000100
^{14}C	5,560 yr	8,020 yr	0.212
^{60}Co	5.24 yr	7.56 yr	0.000858
^{90}Sr	27.7 yr	40.0 yr	0.0068
^{137}Cs	26.6 yr	38.4 yr	0.00994
^{32}P	14.3 d	20.6 d	3.43×10^{-6}
^{51}Cr	27.8 d	40 d	10.6×10^{-6}
^{59}Fe	45.1 d	65 d	19.8×10^{-6}
^{75}Se	127 d	183 d	71.2×10^{-6}
^{85}Sr	64 d	92.3 d	40.7×10^{-6}
$^{99\text{m}}\text{Tc}$	0.25 d	0.36 d	0.185×10^{-6}
^{131}I	8.1 d	11.7 d	7.95×10^{-6}
^{203}Hg	47 d	68 d	71.6×10^{-6}
^{182}Ta	115 d	166 d	156.3×10^{-6}
^{192}Ir	74.5 d	107.5 d	106.5×10^{-6}
^{198}Au	2.7 d	3.9 d	4.00×10^{-6}

and that of ^{238}Th are relatively short. From Table 4, 1 microcurie of ^{238}U requires 2.92 grams of metal. Hence, a 10-pound block of ^{238}U (4,530g) has an activity of about 1.55 mCi, as does its descendent ^{234}Pa very shortly after preparation of the block.

The uranium block very effectively absorbs virtually all the gamma rays from the ^{234}Pa . In addition, the uranium is normally in turn enclosed in a lead housing to supplement its shielding; the housing is more than adequate to shield against *curie* quantities and readily absorbs the gamma rays of the ^{234}Pa .

Secondly, could one have a rich radium ore? Natural radium is always found in equilibrium with ^{238}U ; consequently they have the same radioactivity. As a result of this fact there are always 2,920,000 grams of ^{238}U associated with 1 gram of radium (Table 4). It is therefore clear why the Curies performed a backbreaking job to isolate a small amount of radium salt from uranium ore pitchblende, since one gram of radium is distributed in over three tons of uranium!

Thirdly, as radium decays both radon and helium gases (from the alpha rays) are produced. How great a pressure is produced in a sealed 10 mg container?

First consider the radon. At equilibrium 10 millicuries of radon exist, or 0.0635 micrograms. The volume of this at 0°C, 760 mm Hg is only 0.0000064 ml. This is insignificant as a source of pressure.

What of the helium produced? Assume all 5 alpha disintegrations occur for a full 30 years. Hence, 5 helium atoms are released for each radium disintegration, or each second a 10 mg radium source in equilibrium produces $5 \times 10 \times 37$ million = 1,850 million helium atoms. In 30 years, or 0.946 billion seconds, 1.75×10^{18} atoms of helium are produced. Since there are 6.02×10^{23} atoms in a gram atomic weight, this is equivalent only to 0.065 ml at 0°C and 760 mm Hg. Conceivably, this 1/16 ml at 0°C and 760 mm Hg of helium could create significant pressure in a very confined space. If water is also present, its decomposition products from alpha irradiation could further aggravate the problem. It is generally believed, however, that an initially dry source is unlikely to build up dangerous pressures in periods of less than 100 years, and radium sources are normally re-encapsulated much sooner than this.

Radium and Radon Sources

We now consider the application and nature of radium sources as well as some problems associated with their use.

COMMON APPLICATIONS.^{2, 3} Radium sources are used in three basic ways: interstitially, alongside skin and mucosal surfaces, and in body cavities.

When the diseased area is accessible sources may be inserted directly into tumor and surrounding tissues (interstitially). Applications include both skin cancer and primary and metastatic tumors of the oral cavity, vulva, anus, and lymph nodes. Radon seeds are sometimes employed instead of or in addition to needles (see below).

Surface applicators or molds hold radium alongside tumor tissue. The object is to accurately position the multiple sources so as to deliver reasonably uniform dosage to the tumor. Although skin lesions are most easily treated with molds, ingenious techniques have also been worked out for accessible tumors in body cavities.

Intracavitary treatments are perhaps the most frequent brachytherapy application. Of these, the commonest sites treated are cervix and uterine corpus.

DESCRIPTION OF SOURCES. Four types of these sources exist: radium needles, tubes, capsules, and radon seeds. In addition, tubes are often

used in special applicators to obtain desired dosage distributions, especially for therapy of cervix carcinoma. Needles and some tubes have holes at one end for strings with which to withdraw sources after treatment. A tube with such a hole is usually called a "capsule."

Modern radium sources are doubly sealed in platinum-iridium alloy containers for mechanical strength and proper confinement of radon gas. Radium sulphate salt in an inert filler is first sealed into a small cylindrical gold foil "cell." The filler assures uniform radium distribution in the cell. The cell is then inserted into the main tube or needle and sealed again. The procedures involve considerable hazard as well as skill, and a major part of the cost of radium sources arises from their encapsulation, testing, and standardization.

Figure 3 shows cross-section sketches of some radium and radon treatment sources. Needles [Fig. 3(A)] are usually thinner than tubes or capsules because they must be inserted directly into tissue. Usually the cells are about 0.5 to 1.0 mm in diameter and the walls, 0.5 mm thick. (A common total diameter of needles is 1.65 mm.) The "active length" includes the cells themselves, normally 1 cm long each in needles. The cells may be 0.25, 0.33, 0.50, 0.66, or 1.0 mg each in strength, depending on the therapist's choice of treatment protraction. Usually two needle intensities are employed, such as 0.5 and 1.0 mg or 0.33 and 0.66 mg/cm cell strengths, for reasons to be discussed below.

Two special types of radium needles merit special mention. The first is the "Indian-club" needle. This is deliberately loaded non-uniformly, with a stronger cell at the *point* end. Such needles deliver more intense irradiation to inaccessible deeper tissues which cannot otherwise be adequately treated. The second is the "seed" needle, used to deliver extra dosage to small areas where a small low activity needle is desired.

Tubes and capsules [Fig. 3(B)] differ from needles in two important ways. First, their cell strengths are much greater (say, 5 or 10 mg activity in 12 mm active length) since they are positioned less close together and more remote from the tumor. Second, they may be supplied with thicker walls (1.0 mm Pt) because they are not used interstitially. This makes them more rugged and durable, an important consideration because platinum alloys are relatively soft.

"Radon gold seeds" [Fig. 3(D)] are used in interstitial therapy. They are made of 0.7 mm diameter gold tubing with a 0.1 mm bore. The tubing is filled with radon gas pumped from an acidulated radium bromide solution contained in a special apparatus called a "radon plant." Blunt-edged cutting pliers are used to cut off and pressure-seal tiny gold "seeds" containing Rn gas [Fig. 3(C)]. In practice gold seeds are made about 2 to 3 mm long and prepared to have about 1 mCi radon strength on the day they are inserted into tumor. The filtration

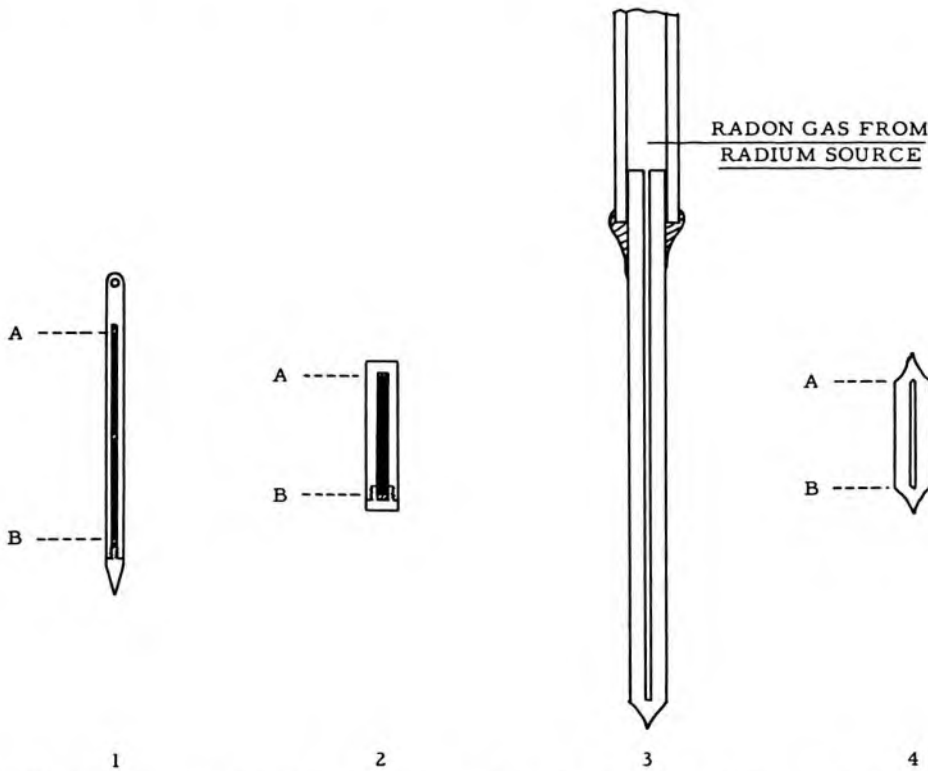


Fig. 3. Radium and radon sources: basic construction (active lengths between A and B). 1. Section of radium needle with two 1 cm long cells. 2. Section of radium tube with enclosed sealed cell. (Same scale as needle). "Capsules" are essentially tubes with an eyelet for attaching string to permit withdrawing source after treatment. 3. 0.7 mm outside diameter gold tube in position to receive radon for making radon seeds (4) (greatly enlarged).

is only 0.3 mm Au rather than 0.5 mm Pt or Au; this represents a compromise between full filtration and requirements of cost and ease of insertion. In using radon gold seeds leakage is often a problem. The author has on one occasion received a shipment of nine seeds, three of them "duds." Consequently, extra seeds should generally be ordered. In addition, it is wise to receive them a day or two in advance. Those destined to leak will ordinarily do so the first day or two. Radon seeds are readily checked with a cutie pie (a portable survey ionization chamber instrument, Chap. 13) and a one or two milligram radium reference source, using a standardized geometry. The short half-life of radon must be kept in mind when ordering and assaying radon seeds.

FILTRATION. Effectively removing beta rays from elements in the radium cascade requires 0.5 mm of platinum or its equivalent. A perhaps more important reason to provide this much filtration is that this thickness of platinum alloy is a minimum to assure adequate mechanical strength. Greater thicknesses are, unfortunately, impractical with needles, which must be forced directly into tissue, but tubes

are made with up to 1 mm Pt wall thickness. Radium dosage tables and charts are generally based on sources with 0.5 mm Pt filtration; 1.0 mm Pt filtration yields intensities about 12 percent lower.

Other materials are occasionally substituted for platinum, the most common being gold alloy, which is equivalent to platinum in absorption. In some applicators silver or even lead can be used; such materials require about 0.2 mm thickness for every 0.1 mm of platinum they replace. Brass, monel, and stainless steel have also been used for capsule materials; they generally require about 0.3 mm thickness for every 0.1 mm of platinum they replace. In the past rather thin radium needles made with monel metal sheaths have been employed. These have very little filtration and do not generally meet modern encapsulation standards.

SAFETY PROBLEMS PECULIAR TO RADIUM SOURCES. The radiochemical nature of radium creates its own unique safety problems. They arise from two causes: radium salt is toxic, and radium produces radon gas. Spread of the salt is a very serious development. Contamination is persistent because of the long half-life of radium. Furthermore, in common with other heavy elements, radium is deposited in bone, where it tends to remain fixed while its alpha rays produce great biologic injury. More than 1 or 2 micrograms of "fixed" radium is a potentially serious health hazard. The medicolegal consequences of radium contamination can be quite serious, involving many thousands of dollars cost in shutdown, isolation and decontamination of the involved area, as well as for personnel medical studies. Radon gas leakage is important both in itself and as an indication of potential leakage of radium salt. Historically, uranium miners may acquire lung ailments from inhalation of radon gas, which leaves "active deposits" of radon daughter nuclides in the respiratory tree. *These considerations make it necessary to ventilate radium storage and handling areas well and to discard or reseal leaky sources.*

Radium salt problems fortunately arise relatively infrequently. The following horrible examples however illustrate what potentially can happen:

1. Radium sources are transferred by a pneumatic tube system. A leaky source releases salt, which is efficiently distributed over a wide area by the pneumatic and ventilating system.
2. Radium sources are kept in vertical holes in a heavy drawer. Careless source replacement and closing of the drawer shears sources in half.
3. A source is lost in a patient's room, swept up by the janitor, and incinerated.

Radon leakage is much more common than salt leakage. Most

TABLE 5. STEPS IN THERAPEUTIC USE OF RADIUM

1. Storage	Should be in safe, with adequate lead shielding.
2. Preparation	Tools: long forceps and source holder. Shielding: ell plus lead bricks.
3. Transfer	Proper shielded container, which can be moved while kept away from personnel.
4. Operating Room	Proper shielding and remoteness during storage; proper procedures in handling during insertion.
5. Patient handling	Isolation to reasonable extent in recovery room and patient's room; proper instructions to nurses and residents.
6. Retrieval of radium	Well instructed resident staff.
7. Transfer back	COUNT SOURCES.
8. Temporary storage	Adequate shielding when returned material is OUTSIDE SAFE for required period of storage.
9. Cleaning, return to safe	Same as 2 above.

local and state radiation safety codes require annual or semiannual testing of all radium. Older sources are especially likely to prove leaky. Of a group of ten old monel metal sources tested by the author, eight were found to have excessive radon leakage; of these, three were grossly leaky, with virtually no seal of radon at all.

In Chapter 14 we consider radioactive safety problems in some detail. However, it is useful to itemize the steps in the therapeutic use of radium (Table 5), in every one of which care must be exercised to assure safety of personnel, patients, and innocent bystanders.

RADON LEAKAGE TESTING. Many methods have been proposed to check radium sources for radon leakage. Basically all tests seal the source in a container for a few days with a material which hopefully traps radon gas. The source is then removed and the accumulated radioactivity is measured (primarily as ^{214}Pb and ^{214}Bi gamma rays) to evaluate the original leakage.

Two basic pitfalls exist in such procedures. First, very few materials trap radon well. Balls of cotton have been used, but they are virtually no better than the container itself and much poorer than a rubber stopper containing graphite. Second, any container leak results in substantial loss of radon gas before active deposits are formed. The diffusion of radon is a serious source of error, as it gives a false indication of safety.

Hale employs activated charcoal in water, using a sealed test tube; the accumulated activity is later easily measured in a well

counter.⁴ Rozenfeld has shown the water is unnecessary and actually harmful.⁵ Our own work indicates at least 2 ml of dry, activated charcoal is desirable in the Hale method to minimize errors from radon leakage during the storage period. So-called "charcoal filters" of cigarettes have proven worthless for this purpose and should not be used.

We test our sources every six months, four sources to a well-sealed test tube containing 2 ml of activated charcoal, usually stored over a weekend. The resulting activity is then measured in a well counter. A reference National Bureau of Standards sealed radium standard is used to obtain a quantitative measure of accumulated activity; this measure is then used to estimate actual rate of radon leakage. Sources are checked individually if the group leakage activity is high, and leaky sources are discarded.

How much leakage is permissible? We are suspicious of any atypically high leakage and reject sources whose total measured leakage approaches 0.005 microcurie. This is consistent with United States Atomic Energy Commission standards for radioactive source wipe tests. (See Chapter 14).

SOURCE ABUSES. Some older sources do not meet modern standards of construction. Many were made with only a single rather than double seal against radon gas leakage, and with moisture inclusions. Also, radium salt is free to move in some cells, causing uncertain source distribution. Thin-walled monel needles have relatively thin walls; in addition to delivering excessive beta irradiation, these sources often leak badly in use.

Some older needles have very great cell activities—up to 10 mg/cm. Therapists have in the past used these for high intensity radium therapy. More protracted therapy is generally preferred nowadays to permit maximum recovery of healthy tissues, and needle strengths of 1 mg/cm maximum are generally preferred.

Radium sources, especially needles, must be handled with care because they are fragile and easily bent or otherwise damaged.

In general newer double-seal sources, of appropriate cell strengths and filtration, are preferred. Older sources may be prone to leakage and poor in other respects as well. Re-encapsulation is recommended wherever a source is substandard in any major respect.

Other Brachytherapy Sources

The problems peculiar to radium have led to a search for suitable substitutes. These fall into two broad groups: those of relatively long and short half-life (Table 6).

TABLE 6. SOME BRACHYTHERAPY RADIONUCLIDES

Isotope	Half-Life	Major Photon Energies in MeV	Specific Gamma Ray Constant Γ^*	mm Pb [†] HVT	Physical Form	Applications
Radium ($\frac{1}{2}$ mm Pt)	1,622 yr	0.053 to 2.43	8.25	16 [‡]	Ra salt in Pt-Ir needles and capsules	Interstitial, surface, and intracavitary use
⁶⁰ Co	5.24 yr	1.17 and 1.33	13.4	13	Fine Co wire in Pt-Ir or st. steel tubes	Interstitial, surface, and intracavitary use
¹³⁷ Cs	26.6 yr	0.662	3.3	5.5	Salt sealed in Pt-Ir	Interstitial surface, and intracavitary use
¹⁹⁸ Au	2.7 d	0.41	2.5	3.0	Small cylindrical pellets	Interstitial use with special gun
¹⁹² Ir	74.4 d	0.283 to 0.613	5.5	2.4	Small pellets in strings of nylon tubing	Interstitial afterloading technique
¹⁸² Ta	115 d	0.05 to 1.24	6.1	9.8	Thin wire in Pt sheath	Interstitial use

* Roentgens delivered at 1 cm distance per millicurie-hour using tiny source. 0.5 mm Pt filtration assumed for radium.

† Broad-beam value.

‡ High value, after heavy filtration.

LONG-LIVED RADIONUCLIDES. Cobalt-60 and cesium-137 are the most popular of these materials used to date. Cobalt-60 comes in wire form. It is plated with nickel because otherwise the wire may shed radioactive dust. The half-life of cobalt-60 is only 5.24 years, so source activity corrections are required every three months to assure a ± 1.5 percent accuracy. However, the metallic form makes cobalt-60 quite versatile.

Cesium-137 has recently become available as a direct substitute for radium. Its 26.6 years half-life makes decay correction once a year sufficient, an advantage over cobalt-60.

Both these materials are not concentrated by bone and do not release any radioactive gas. They are hence much less toxic than radium. In addition, the gamma rays from cesium-137 are somewhat softer than those from radium, facilitating radiation shielding both in storage and handling of sources.

SHORTER HALF-LIFE MATERIALS. Many radionuclides are available with interesting properties, but gold-198, iridium-192, and tantalum-182 are most commonly employed. All these are rather costly since their half-lives are relatively short (2.7, 74.4, and 115 days, respectively), and sources can therefore be used only a limited number of times. However, total expense of treatment includes handling and other costs as well as those of sources, so these materials may well have greater use in the future.

Gold-198 grains or seeds are made with a special shielded unit which cuts and inserts them. The radioactive gold wire used is made from ordinary gold wire which is irradiated in a nuclear reactor. Gold grains are used in much the same way as radon seeds.

Iridium-192 is used as small pellets in nylon tubing, spaced 1 cm apart. These nylon tubes are threaded through tumor tissue and are especially useful in planar implants for head and neck work. Because of its longer half-life, ^{192}Ir is removed after an appropriate time (usually several days).

Tantalum-182 is used as a platinum-coated thin wire. To treat bladder cancer, it has been used in the form of hairpin-shaped wires at the end of strings; when treatment is completed the wires are removed through the urethra by simply pulling the strings.

BRACHYTHERAPY DOSIMETRY PRINCIPLES

The therapist faces two tasks in planning brachytherapy treatment. The first is to arrange a reasonable number of sources so as to achieve acceptably uniform dosage to the tumor bed. The second is to determine the millicurie hours then required to obtain the desired

TABLE 7. EXAMPLES OF CLINICAL APPLICATION OF RADIUM DOSAGE DATA

A. Planar data	<ol style="list-style-type: none"> 1. Superficial lesions: plaques. 2. Mucosal lesions: molds.
B. Volume data	
1. Planar	1. Reasonably flat lesions treatable with one or two plane array of sources.
2. Volume	2. Lesions where length, width, and thickness are of same order of magnitude (tongue, lymph nodes).
C. Other data	
1. Charts of dosage distributions around pelvic radium sources	1. Planning of treatment of cervix and uterine carcinoma.
2. Linear source data	2. Basic to computer calculations of dosage when positions of radium sources are known.

tumor dose in rads. In practice the tumor location may present a simple or complex geometry (Table 7). Some applications (groups A and B) involve simple anatomic geometry. Group C refers to some locations where tumors are usually not readily approximated by cylindrical and planar source distributions. These include the uterus and uterine cervix, the most frequent brachytherapy sites.

Various brachytherapy procedures and dosimetry systems have been developed, and the reader is referred to the references for detailed information. In this section we shall limit our discussion to brachytherapy dosimetry principles and concepts, presented in the following order:

1. Dosimetry—single tiny source
2. Dosimetry—single line source
3. Application of multiple sources
4. Multiple source dosage data.

Dosimetry—Single Tiny Source

By a "tiny source" we mean a small one compared with distances of interest (for example, less than 1 mm in diameter). With such a source an inverse square relationship applies closely, and we may write:

$$D = \Gamma \frac{M}{d^2} \quad (12-4)$$

where D is the exposure in roentgens at a given location
 d is the distance from the source to this location
 M is the millicurie-hours of radionuclide disintegrations.

The term Γ is the specific gamma-ray constant for the particular gamma emitting radionuclide employed; it is discussed below. In practice this equation is reasonably accurate for actual sources when the treatment distance is at least 5 times greater than the largest source dimension.

Note (12-4) does not consider attenuation of gamma rays in their travel to the location of interest. This effect is also discussed below.

THE SPECIFIC GAMMA-RAY CONSTANT Γ . Γ is the exposure in roentgens to a point 1 cm distance in air from a tiny source during 1 mCi-hr of radionuclide decay. Most radionuclides are used without special filtration. However, radium sources require encapsulation to confine the radon gas and radium salt. Consequently the Γ value for radium (or radon) is specified for a source filtration of 0.5 mm platinum. Table 6 presents data for radium and five other useful gamma emitters.

The gamma constant for a radionuclide can be measured directly, but this is an experimentally complex procedure. Calculation is possible if the gamma spectrum is known, using available air $(\mu/\rho)_{en}$ data and the definition of the roentgen.⁶

SUBSTITUTION OF OTHER GAMMA-RAY EMITTERS. The available brachytherapy dosage tables generally refer to radium with 0.5 mm platinum filtration. When radium is used with other filtration such as 1.0 mm Pt, the output is reduced roughly 2 percent for each added 0.1 mm of Pt (12 percent lower for 1.0 mm Pt).

When other radioactive materials are used the output per millicurie-hour depends on the particular gamma-ray constant. In general, the ratio of Γ values is involved:

$$D = D_0 \frac{\Gamma}{\Gamma_0} \quad (12-5)$$

where D_0 = the dosage per millicurie-hour of radium

D = the dosage per millicurie-hour of the alternative radionuclide

Γ_0 = the specific radium gamma-ray constant

Γ = the specific gamma-ray constant of the new radionuclide.

Alternatively one might obtain the required millicurie-hours to deliver a desired tumor dosage using radium (M_0) from standard radium tables and wish to determine that needed using the new radionuclide (M). The relationship is simply:

$$M = M_0 \frac{\Gamma_0}{\Gamma} \quad (12-5')$$

Note that (12-5) and (12-5') usually apply with adequate clinical accuracy to all practical line sources and groups of sources as well as the tiny ones considered above.

Single Linear Source

Treatment sources are usually of significant size compared with the distances involved. For example, sources range in active length from about 10 to 15 mm for tubes and capsules to up to 5 cm for needles, as compared with treatment distances of 0.5 to 3 or 4 cm. Under these circumstances (12-4) is inapplicable. Two factors contribute to the departure from the inverse square law: geometry and oblique filtration.

GEOMETRY AND OBLIQUE FILTRATION—EFFECT. To illustrate how these factors operate, Figure 4 compares a "point" (tiny) radium source with a surrounding spherical shell of platinum alloy 0.5 mm thick with a straight source 5 cm long, surrounded by a cylindrical platinum alloy shell also 0.5 mm thick. Now, consider the relative dosage levels at points A and B, 2 and 4 cm from the source centers.

With source (A) of Figure 4, B receives $(2/4)^2 = 1/4$ as much dosage as A, since (12-4) applies. Also, the beam traverses the platinum perpendicularly, and exactly 0.5 mm thickness is always traversed by emitted gamma photons.

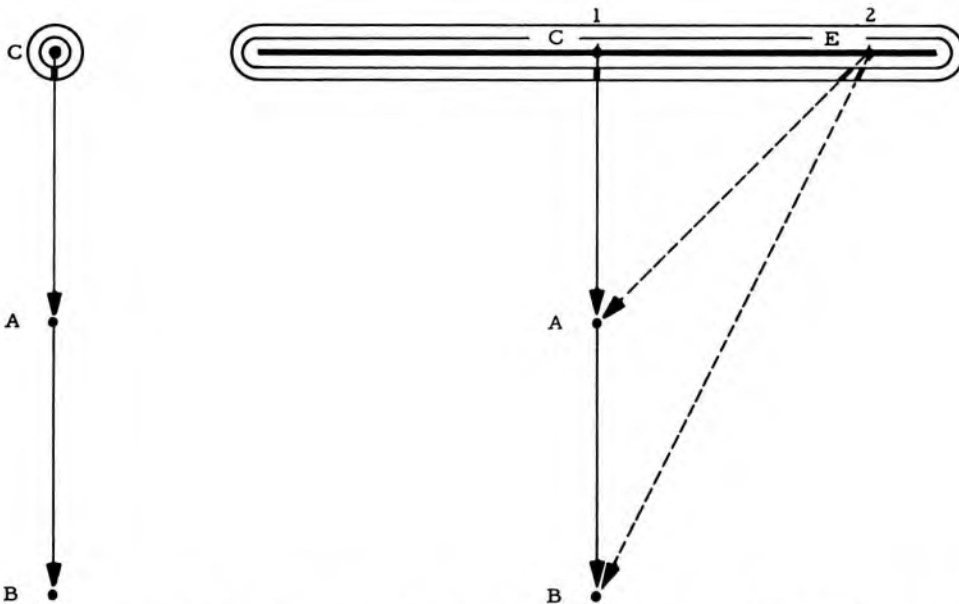


Fig. 4. Geometry and filtration effects. Left. Point, or tiny, radium source. Inverse square law applies accurately, and filtration is the same for rays emerging in all directions. Dose at B is 0.25 times that at A. Right. Five cm long straight source, same radioactivity and filtration as at left. With the straight source, dose at B is 0.317 times that at A, a greater fraction than at left.

In source (B), Figure 4, this simple situation is true only for a small part of the radium in the center (1). However, all other parts of the linear source are some distance from the line CAB along which the inverse square law applies. Consider the contributions from location (2), 2 cm from the center. Due to beam obliquity, distance AE is $2\sqrt{2}$ cm, distance EB, $2\sqrt{5}$ cm. Were the filtrations the same the ratio of doses to B and A would be 0.4, *not* $\frac{1}{4}$ as called for by (12-4). In general, the dose from a source falls off *less* rapidly with perpendicular distance from the center than predicted by the "inverse square law."

A second new factor, oblique filtration, operates in Figure 4(B). Note that indicated rays leaving (2) traverse the platinum obliquely, unlike (1) in Figure 4 (B) and all those in Figure 4 (A), so filtration is substantially greater. In the example shown, the oblique ray EA traverses 0.707 mm Pt, ray EB, 0.56 mm Pt. Hence oblique filtration also reduces the dosage delivered by any line source.

LINEAR SOURCE DATA—EXAMPLE. Quimby computed linear source dosage data several years ago; these have more recently been recalculated by Greenfield to produce values which lie on smoother curves. Using Greenfield's data,⁷ we have prepared Figure 5, which compares the dosage variation around a 4 cm active length 4 mg radium source (0.5 mm Pt) with that around a point source of the same strength and filtration. Note the computed point source data are reduced by the proper factor of 4 when the distance is doubled from 1 to 2 cm. However, the corresponding reduction factors for the linear source near the center and end are only 2.80 and 2.33, respectively. Also, the outputs are consistently lower for the needle than for the point source of the same activity, at any given distance from the source. This is primarily because the needle is essentially a linear array of a large

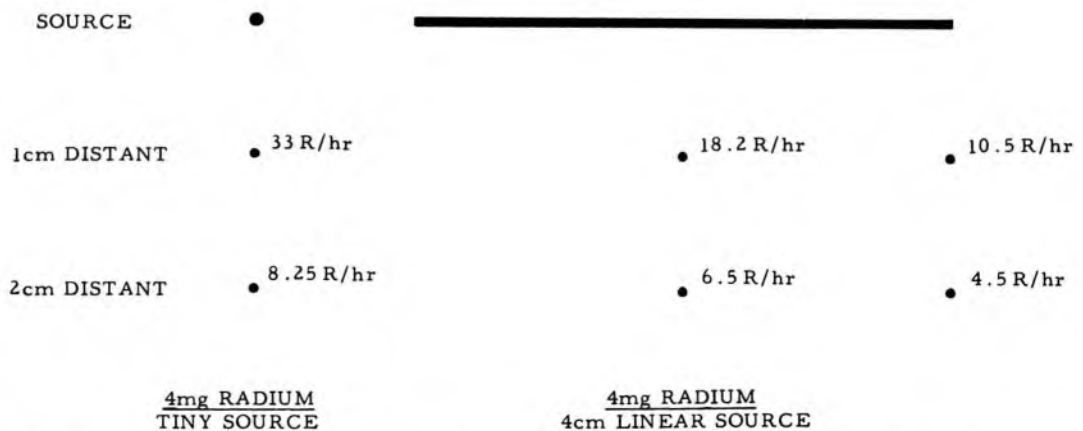


Fig. 5. Comparison of exposure rates at 1 and 2 cm from two sources, both 4 mg of radium, with 0.5 mm platinum filtration. Left. Point source. Right. Linear source 4 cm long. This example illustrates principles of Figure 4.

number of tiny sources whose distances from the points in question are all greater than 1 or 2 cm, and secondarily because of oblique filtration.

It is clear from this discussion that one cannot usually assume radium sources to be point sources in brachytherapy without significant errors.

Multiple-Source Data

Table 7 indicates the various types of data in common use and the clinical situations in which they are most applicable. The best planar and volume procedures and data are those of the Paterson-Parker (Manchester) system.⁸ Greenfield's linear source data⁹ are most often combined with roentgenographically verified source positions to calculate intracavitary dosage levels.

All dosage data relate the mg-hours of radium to roentgen exposure. In the Manchester system, data are presented as mg-hours per 1,000 R at relevant locations; in the Quimby and Greenfield data, as R per mg-hr. All are for 0.5 mm Pt filtration radium sources. Corrections for other filtrations and nuclides have been considered previously.

In our discussion below, we shall first briefly discuss the basic question of source arrangement. We then describe the Paterson-Parker tables and finally consider intracavitary therapy dosage procedures.

SOURCE DISTRIBUTION. Paterson and Parker designed radium source arrangements which assure uniform *tumor dosage* distribution (say to ± 10 percent) at distances more than 0.5 mm from individual radium sources. The source arrangements are too complex to describe in detail here, but follow three basic principles:

1. A great many sources are used, with a variety of strengths.
2. Much of the radium is located around the periphery of the lesion. This produces a crossfire or "multiple portal" effect. The exact arrangement depends on the geometry involved.
3. For larger lesions, additional sources are placed appropriately at interior locations to supplement the dose from peripheral sources.

Undoubtedly, this is the best way to do surface and interstitial radium therapy, and major cancer centers all employ these basic principles. However, many hospitals with fewer facilities also carry out radium therapy, usually with far fewer sources and a limited variety of strengths. (By source strength in this context is meant the mg/cm active length of needles. These come in 0.25, 0.33, 0.66, and 1.0 mg/cm strengths.) Another difficulty is that often it is difficult to insert needles according to the Paterson-Parker rules, and a simple uniform source arrangement is more feasible. Quimby therefore

TABLE 8. THREE SYSTEMS OF SOURCE ARRANGEMENT IN BRACHYTHERAPY

Item	Paterson-Parker	Quimby	Modified Quimby
Source Arrangement	Periphery and core with specified rules	Uniform spacing	1. Half-strength sources in center 2. Cross ends
Dose Distribution	Uniform, $\pm 10\%$ in specified locations	Center "hot," edge "cold"	Compromise between the other two
Millicurie-hours treatment to obtain same minimum tumor dose	Lowest possible	Greater	In between the other two

developed dosage data for the less desirable but widely employed simple uniform distribution of sources. In such an arrangement the center tends to be overtreated.

Over the years some radiotherapists decided in collaboration with physicists that a compromise between the two basic approaches was required, and the "modified Quimby" type distributions resulted. These use uniform source distributions but with the following modifications that essentially incorporate the principles of the Paterson-Parker system:

1. Half-strength sources are used for interior locations to make the central and peripheral doses more nearly equal.
2. Ends of needles are crossed with others where possible, to increase the dose at the ends of the array.

Table 8 summarizes relevant aspects of the three systems.

Very high doses are, of course, delivered close to radium sources. Remarkably uniform dosage distribution, however, is achievable a few millimeters away from them. Figure 6 shows computed dosage distributions for two $3 \times 3 \times 3$ cm cylindrical implants, taken from Fletcher.¹⁰ While tissue near to needles is treated very intensively, large surrounding areas are relatively homogeneously treated.

CORRECTIONS FOR PATERSON-PARKER DATA.¹¹ Paterson-Parker tables give data in roentgens rather than rads. In addition, they are computed assuming the radium constant is 8.40, rather than the newer figure of 8.25 R/mg-hr at 1 cm. Paterson hence suggests a correction factor of 0.95 in use of his data. (0.965 is for the \bar{f} value, and the rest for the ratio of the Γ values.)

Shalek has suggested that significant errors arise from tissue attenuation and oblique filtration.¹² All published radium dosage data ignore tissue attenuation; fortunately, the errors introduced are mini-

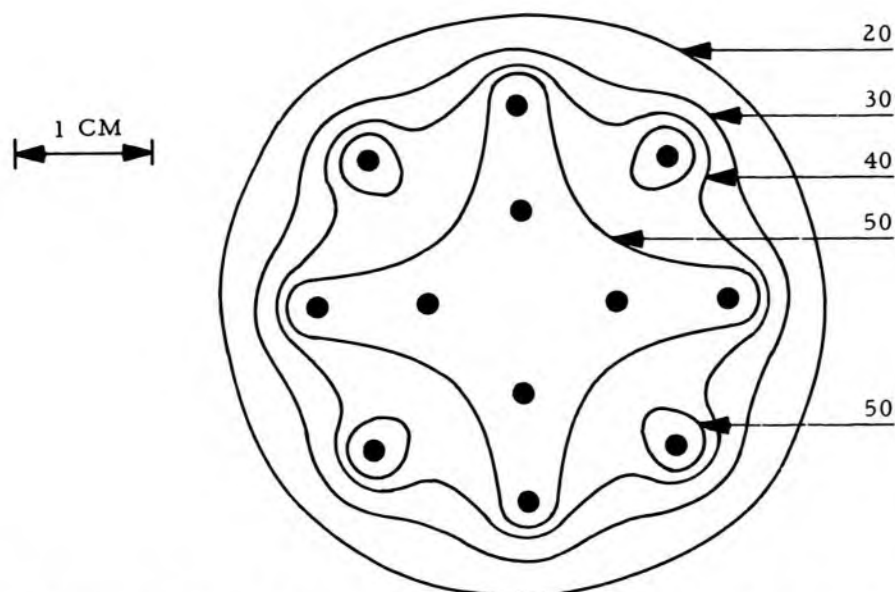


Fig. 6. Computed midplane dosage distribution for a $3 \times 3 \times 3$ cm needle implant, both ends uncrossed. All needles are shown as black dots, and are 0.33 mg/cm strength. The single dosage figure obtained from Paterson-Parker data is 45.4 rads. (Redrawn from Fletcher.¹⁰)

mal for two reasons. First, distances are small, and second, there is significant contributory scatter which compensates for attenuation to a considerable extent. Shalek estimates the error from ignoring tissue attenuation to be about 2 percent/cm distance in needle implants. He finds a combined correction factor of 0.90 reasonably accurate for interstitial therapy applications of Paterson-Parker data. Of course, if radium sources of different filtration are used, appropriate corrections must also be made.

PATERSON-PARKER DATA. There are four types of Paterson-Parker tables, pertinent to four radium treatment situations. These will now be briefly described.

Surface applicator data are intended for skin and mucosal surface lesions of small to moderate areas, which can be reasonably approximated using plaques or molds. Data are given for areas up to 100 cm^2 , and distances of 0.5, 1.0 etc., to 5.0 cm. This table is supplemented by another for larger areas up to 400 cm^2 , for use in radium treatment of locally recurrent skin lesions from breast carcinoma.

Two types of volume implant data are given, pertaining to planar and volume implants. The former table is used in situations in which the tumor distribution is relatively flat. The table dosage value refers to the minimum dose in roentgens 0.5 cm on either side of the plane of the radium. For thicker tumors, two parallel planes of radium sources may be employed with appropriate increases in the total required radium.

TABLE 9. PATERSON-PARKER RULES FOR SURFACE APPLICATORS

A. Separation of active ends of sources should not exceed h , the treatment distance

B. Circles—diameter d .
 Arrange radium on outer circle, with inner circle diameter $d/2$, and a central spot. Percentages of total strength indicated at right.

d/h	1 to 3	3 to 6	6	7.5	10
outer circle	100	95	80	75	70
inner circle	0	0	17	22	27
center	0	5	3	3	3

C. Ellipses—consider as circles if eccentricity is small.

D. Rectangles— $a \times b$.
 a , short; b , long side

a less than $2h$. All radium on periphery.
 a more than $2h$. Add extra lines of radium parallel to longer side, $2h$ apart.
 For *one* extra line, use $1/2$ mg/cm as edge.
 For 2 or more extra lines, use $2/3$ mg/cm as edge.

E. Elongation—factor b/a :
 increase mg hr by:

2	3	4
5%	9%	12%

F. Concave and convex surfaces—compute mg hrs from actual *treated* area, but distribute it over the actual *mold* area.

In many tumors a more nearly cylindrical volume is required to encompass the tumor. In such implants, the system provides a table for obtaining the required mg-hr for different volumes enclosed by the radium.

It must be stressed that the Paterson-Parker tables constitute only one half of a unified system. The source distribution rules must be scrupulously applied, or incorrect dosage values will be computed. A concise summary of these rules is given in Tables 9 and 10, from Meredith.¹³ The reader is referred to the references for tabular data and illustrative examples.

INTRACAVITARY DOSIMETRY.¹⁴ Tumors in some anatomic sites are not amenable to treatment by either surface or interstitial therapy. The uterine cervix and corpus are major examples. A wide variety of methods have been developed for treating tumors in such locations. All, however, employ essentially the following basic procedures:

1. Use charts to decide on the desired source strengths and locations.
2. Insert radium with appropriate applicators, spacers, packing, sutures, etc., to assure reproducible geometry.

TABLE 10. PATERSON-PARKER PROCEDURES—
VOLUME IMPLANT PRINCIPLES

A. Planar implants

1. Basic: 1 cm thick slab treated, 0.5 cm either side of single plane containing needles.
2. Two-plane: for planar blocks up to 2.5 cm thick, use two equal parallel planar distributions of radium. Increase total radium appropriately.
3. Surface applicator data may be used to compute planar implant dosage.

B. Volume implants: sphere, cube, or cylinder; more than 2.5 cm thick.

1. 75% of radium peripherally, 25% more centrally.
2. Cylinder
 - (A) Belt: 50% of total radium with at least 8 needles.
 - (B) Ends: 12½% each end.
 - (C) Core, rest: 25%, with at least 4 needles.
 - (D) For each uncrossed end, reduce the volume figure employed in table by 7.5%.
 - (E) Elongation:

Length/diameter	1.5	2.0	2.5	3.0
Increase mg hr by	3%	6%	10%	15%

3. *Verify the source locations and change if necessary.*

- a. Use a dosimeter probe to verify that the maximum rectum and bladder dosage is not excessive.
- b. Radiograph the patient, usually with contrast media, using perpendicular views and an opaque circular ring for magnification correction. From this, determine minimum source distances from vulnerable organ locations.
- c. Use linear source tables to verify dosage to important areas. These include points A and B as well as the nearby rectum and bladder walls.

OTHER RADIONUCLIDE DOSIMETRY

In addition to brachytherapy, radionuclides are used in at least three other ways for treating malignant disease. These are in teletherapy of *tumor volumes* using intense gamma sources, in local treatment of *organs* using true solutions, and in localized treatment of *surfaces* inside and outside the body with beta-emitting materials.

Teletherapy

Various names have been applied to machines used in teletherapy. They include bomb, pack, telecurie, and other terms. A single source

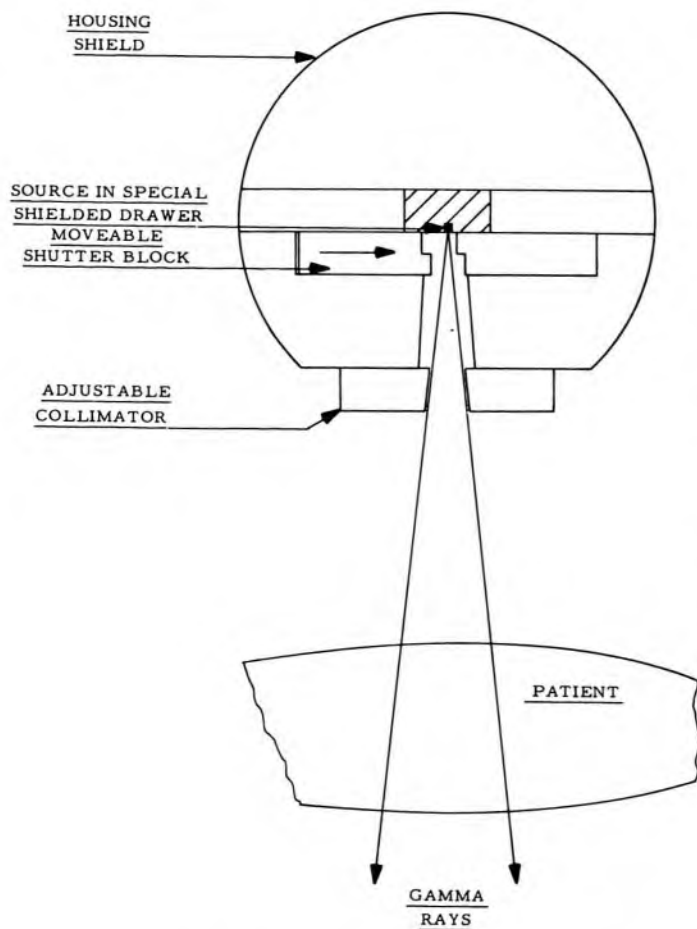


Fig. 7. One type of teletherapy assembly, showing source capsule in a housing shield. A moving shutter block below the source exposes the patient as desired to the gamma rays. The field size is determined by the adjustable collimator below the housing. Capsule self-absorption reduces the output below theoretical values. Contributory scatter from the source, housing, and collimator add significantly to the direct radiation. These effects make it necessary to specifically measure teletherapy machine output.

of great radioactivity is enclosed in a suitable, heavily shielded housing; a massive shutter is used to turn the beam on and off, and a collimator system is provided to adjust the treatment field size (Fig. 7).

Both cobalt-60 and cesium-137 sources have been used. ^{137}Cs has two theoretical advantages. First, its half-life is 26.6, compared with 5.24 years for ^{60}Co . Second, its lower photon energy reduces shielding problems, both of the housing and room. ^{60}Co is nevertheless almost universally employed in preference to ^{137}Cs . Its metallic form permits convenient and compact encapsulation. ^{137}Cs is an alkali metal and hence must be used as a salt (usually $^{137}\text{CsCl}$). This fact, in addition to its low Γ value of only 3.3 R/mCi-hr at 1 cm (versus 13.4 for ^{60}Co), results in a very large required source for a given beam intensity, with excessive penumbra. We shall therefore confine our discussion to ^{60}Co teletherapy machines.

BEAM INTENSITY. In most machines the source is small relative to the treatment distance (for example, 2 or 3 cm diameter versus at least 40 cm SSD for cobalt-60). One might expect (12-4) to apply accurately, but in practice, this formula is inaccurate for two reasons.

First, the source has substantial thickness. Consequently, photons originating within the source suffer considerable self-absorption. For a cobalt-60 source 1 cm thick, an average loss of 20 to 25 percent results, so one in general must order additional expensive curies of cobalt-60 to obtain the desired intensity. The second effect is contributory scatter, originating within the source itself as well as in the collimator and other exposed structures. In a particular cobalt-60 teletherapy machine it is estimated that the total beam roentgen exposure in air consists of 80 percent direct, 20 percent contributory scattered radiation. Of course, scatter is softer than the direct radiation, so the actual treatment beam is somewhat softer than would be received from an isolated small cobalt-60 pellet. Beam HVL, however, is apparently not very variable; values of about 10 mm Pb are consistently reported for cobalt-60 units.

It is evident that beam intensity is not easily computed in teletherapy. In practice the central ray intensities at all treatment distances are measured with a suitable thimble chamber, just as with x-ray machines. However, once they are measured, output values may be accurately computed later using a simple correction for decay. This predictability of output is an inherent advantage of teletherapy over x-ray machines.

OUTPUT CORRECTIONS IN TELETHERAPY. Cobalt-60 machine output varies substantially with field size even in air, due to the large amount of contributory scatter present in the beam. This variation is as much as twice that in x-ray supervoltage machines because scatter within the source is substantial.

Of course, field size convention and penumbra problems must be specifically considered, as discussed in Chapter 7.

SOURCE STRENGTH. Because self-absorption is so great in teletherapy sources, they are purchased on the basis of output intensity at a given distance rather than total number of curies. The strength is generally specified in roentgens per minute at a given distance, usually a meter. For example, a strong source could deliver 80 R/m to a 10×10 cm field at 1 meter *installed in the machine*. The source is then called, for short, an "80 RMM" source.

Source costs, of course, depend on the RMM rating. However, the diameter is also involved. Smaller diameter sources contain less material, so a given strength must have more curies per gram specific activity (see Chap. 9). This generally requires more reactor time to achieve, so higher specific activity sources are more expensive. In some high intensity machines the source cost can easily approach that of the basic machine.

True Solution Dosimetry

When true radioactive solutions are administered to a patient, they are generally delivered to all organ systems by the blood. (We assume intravenous administration or oral administration with normal gastrointestinal absorption.) Each organ system will take up, process, and excrete the radioactive material in accordance with its biochemical function and the chemical form of the radionuclide. Dosage calculations are commonly required for two purposes: to verify the safety of a proposed diagnostic radioactive tracer study and to evaluate dosage delivered during radiotherapy to both a tumor and various organ systems. In both applications the concentration and retention of the radionuclide varies greatly and in often unknown fashion, so accurate dosage calculations are extremely difficult to make.

Fortunately, approximate dosage values are often of great value, and estimates are acceptable which involve simplifying assumptions. We shall hence discuss useful approximate formulas for beta and gamma dosage derived by Marinelli and Quimby, but first we shall comment regarding basic problems in the therapeutic use of true radioactive solutions.

THERAPY APPLICATION LIMITATIONS. When radionuclides became generally available for clinical use there were high hopes that true solutions could safely deliver cancer-lethal dosage. It was believed all one had to do was administer a suitable chemical form of the radionuclide. As a result of its avidity for this chemical, tumor tissue would then be selectively injured by ionizing radiation. This result has, unfortunately, not been fully realized; it has been partially achieved to an acceptable degree in only two uses, the treatment of thyroid and certain bone marrow tumors.

The basic problem is the sensitivity of other healthy tissues to ionizing radiation since the blood delivers the radionuclide to the entire body.¹⁵ Even 100 or 200 rads to the entire body produces a severe reaction, but 3,000 to 6,000 rads is usually needed to control a tumor. Thus, the target tissue must receive at least 30 times more dosage than the body generally. Actually, only thyroid tissue exhibits this degree of selective concentration, with inorganic iodine; moreover, it is primarily functioning tissue which thus treats itself; any non-functioning tumor receives minimum dosage. Some thyroid cancers may be usefully irradiated with inorganic ¹³¹I following suitable preparatory medication. The most frequent radiotherapy, however, is for benign disease for which much lower target tissue dosage is required: in hyperthyroidism and palliation of angina disease in euthyroid patients.

Bone marrow selectively concentrates phosphate ion, and $^{32}\text{PO}_4$ is administered to treat polycythemia-vera and chronic leukemia. The degree of concentration is less than one might desire, but, fortunately, the required tumor dosage is also lower than that for most carcinomas.

Dosage evaluation presents difficult problems in this mode of therapy because it is hard to estimate the radioactivity concentration in tissue. Tracer studies before treatment are only a general indication of the uptake of the therapeutic dose. This is because the treatment radioactivity is quite great, and produces almost immediate tissue changes to capillaries and the actively concentrating tumor tissue. Consequently, the therapeutic concentration is generally lower than that predicted from the tracer study. Retention and excretion moreover are also affected in a generally unknown manner. Measurements with a very tiny hole probe scintillation detector may be made to obtain an idea of organ uptake and retention of the radionuclide.

Marinelli and Quimby have developed formulas for computing tissue dosage due to beta and gamma rays for the simplest radioactivity distributions.¹⁶ Rossi has derived more elaborate formulas for less uniform radioactivity distributions.¹⁷ We shall now briefly discuss the Marinelli-Quimby formulas and their application.

MARINELLI-QUIMBY FORMULA: BETA-RAY DOSAGE. The formula is simply stated:

$$D_{\beta} = 73.8 (CT_e)\bar{E}_{\beta} \quad (12-6)$$

where D_{β} = total beta dose delivered in *rads*
 C = concentration of radionuclide in $\mu\text{Ci/g}$
 T_e = effective half-life of the radionuclide
 in the organ in *days*
 \bar{E}_{β} = the average beta energy of the radionuclide in *MeV*.

Two basic assumptions are made. The first is that the radioactive material is uniformly distributed in tissue. The second is that all the beta-ray energy is dissipated in the tissue bearing the radioactivity. This latter requirement is met only when the average beta-ray range is small compared with the relevant tissue dimensions. (See Figure 8.)

In practice these assumptions are only approximated, so calculated dosage values are usually only a rough estimate. Both the T_e and C values, however, are usually evaluated only with difficulty, so more rigorous formulas are usually not justified by the available data.

The formula is quite reasonable. One would expect the rad dose to increase with the mCi-hr (i.e., disintegrations) per gram,

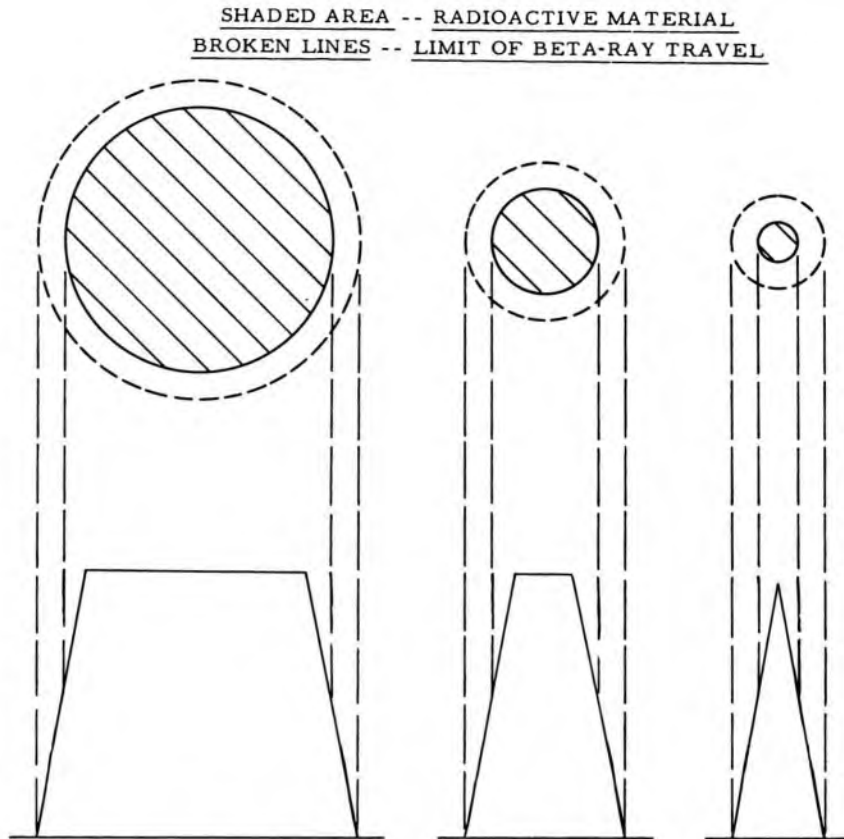


Fig. 8. The effect of tissue dimensions vs. beta-ray maximum range. Upper drawings, radioactivity distribution; lower drawings, dosage distribution. Left. Object large vs. beta particle range. Center. Smaller object. Right. Organ size comparable with maximum beta-ray energy. Note dosage is reduced within organ comparable in size to range of beta particles, and substantial dosage is delivered beyond the organ itself.

which is implied by (CT_e) , and with the available energy/disintegration, implied by \bar{E}_β . C is in microcuries per gram. All these microcuries are destroyed since T_e is the effective half-life; also $T_e = 1.443T$. Hence, (CT_e) implies the millicurie-hours, by equation (9-19). The 73.8 is simply a constant for the units employed.

MARINELLI-QUIMBY FORMULA: GAMMA-RAY DOSAGE. The gamma-ray dosage formula retains the (CT_e) term since millicurie-hours are also involved in gamma dosage; in addition, two other variables, \bar{g} and Γ , are added:

$$D_\gamma = 0.0346 (CT_e)(\bar{g}\Gamma) \quad (12-7)$$

where D = total gamma dose in roentgens (*not rads*)
 C = concentration of radionuclide in $\mu\text{Ci/g}$

$$T_e = \begin{array}{l} \text{effective half-life of radionuclide in} \\ \text{the organ} \end{array} \quad \text{days}$$

$$\Gamma = \begin{array}{l} \text{the specific gamma-ray constant of the} \\ \text{radionuclide} \end{array} \quad R/mCi\text{-hr}@lcm.$$

The \bar{g} term is a geometric constant reflecting the shape and size of the volume containing the radionuclide. Uniform distribution is again assumed. The (CT_e) term again implies the millicurie-hours and hence total disintegrations involved. The $(\bar{g}\Gamma)$ term involves both the distance aspect (\bar{g}) and the gamma constant (Γ). The 0.0346 is simply a constant for the units employed.

Tables of \bar{g} have been prepared for various simple shapes, such as cylinders, spheres, etc. Generally, one is interested in an estimate of whole body dose in using gamma rays because of their great penetration. For the torso a cylinder approximation is often employed, and uniform dosage assumed. For small organs \bar{g} becomes quite small and, if present, beta dosage is the major consideration. For example, about 90 percent of the total dose in thyroid therapy is due to beta irradiation of the organ.

A fuller discussion of the application of these formulas is given in the reference, which also provides illustrative examples.¹⁶

Beta-Ray Therapy of Surfaces

It is sometimes useful to employ beta irradiation locally to treat pathology which is very shallow in depth, of the order of a millimeter or so. This can be on the eye, the skin, or surfaces inside the body.

STRONTIUM-90 EYE APPLICATOR. Several years ago, special ^{90}Sr applicators were developed for treating superficial eye lesions. A thin metal foil confines and protects the active material but still permits beta rays to emerge. The useful radiation is actually from ^{90}Y , the shorter lived daughter of ^{90}Sr . ^{90}Y emits a very energetic beta ray of $\bar{E}_\beta = 0.93$ MeV and $E_\beta = 2.24$ MeV, with corresponding ranges in tissue of 3.65 mm average and 10.5 mm maximum. (See Table 3 in Chapter 9). In commercial units about 50 mCi is used and a factory calibration of the beta dosage provided. Since these units are employed for superficial non-malignant lesions, a rather empirical approach has been generally followed. Johns reports dosage characteristics of some ^{90}Sr applicators.¹⁸

It should be stressed that the dosage distribution is relatively non-uniform across the field of these units, falling off to the sides. Ideally, each unit should be checked by film dosimetry to evaluate the dose distribution across the field. The depth dose characteristic is best

measured with a special extrapolation chamber. Rossi has developed a formula for calculating such data.

SKIN IRRADIATION WITH BETA RAYS. ^{32}P , as a solution dispersed in filter paper, was used by Low Beer several years ago to treat skin lesions. More recently, techniques have been developed to mount ^{32}P uniformly in plastic or as solution in flat, thin-walled containers. Rossi's formula can be used for calculating dosage from such sources if the concentration, thickness, and radionuclide are known.

Haybittle has developed a special arrangement for irradiating more extensive areas of a patient with beta rays from a ^{90}Sr source, but this type of device has not yet been widely used.¹⁹

BETA IRRADIATION OF LESIONS INSIDE THE BODY. We have already referred to the basic difficulty of using true radioactive solutions in the body: they tend to become widely distributed and deliver excessive whole body irradiation. Suspensions or colloidal solutions can be made with particle sizes large enough to prevent their substantial transfer to the general circulation; they are then localized and irradiate more limited areas. This fact has been used to irradiate certain tumor sites.

An interesting application is the preoperative injection of ^{198}Au into a breast with malignant disease. In this situation the colloidal particles are transferred to the regional lymph nodes in much the same way as free tumor cells; the particles are trapped in the nodes and deliver high local dosage to the nodes and hopefully to any small nests of tumor cells they may contain. Estimated average dosage levels of more than 10,000 rads have been delivered in this way, but in-vivo dosage prediction is extremely difficult. A basic difficulty limits the effectiveness of the method: nodes involved with tumor tend to become obstructed, so the ^{198}Au particles are fixed only at the edge of all but very tiny tumor cell masses. Consequently, healthy nodes may be heavily irradiated as well as individual tumor cells, but tumor cells surrounded by other cells are only minimally irradiated. Another application of colloidal radionuclides is flushing of the wound during surgery, when spillage of tumor cells is a possibility. (Thiotepa and other chemotherapeutic agents are also used for this purpose.)

Perhaps the most common therapeutic application of ^{198}Au and ^{32}P colloidal solutions is in the control of fluid accumulation in advanced malignant disease of the pleural and abdominal cavity. Palliation is often achieved by draining accumulated fluid and adding the radioactive colloid in appropriate concentration to irradiate serous surfaces. ^{32}P as chromic phosphate is a pure beta emitter; it therefore offers the advantage over ^{198}Au that systemic gamma irradiation is avoided. Also, its greater particle size minimizes incidental transfer

of radioactivity to the blood. The greater ^{32}P beta-ray penetration can produce some intestinal mucosal irritation, but this is apparently minimal.

The surface dosage from colloidal radiotherapy is very difficult to compute because the colloidal particles deposit out on serous surfaces. Other variables also complicate the situation. For example, the colloidal solution is diluted when more fluid is released from serous surfaces, and loculation of fluid may prevent treatment of some surfaces altogether. In practice, careful preparation and administration of colloid and instructions to patient and nurse are vitally important to assure proper distribution of the material in the cavity. Accurate dosage estimates are extremely difficult to make because of all the variables involved.

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13

Basic Concepts of Protection Against Ionizing Radiation

Grubbé identified injury from x-rays during the summer of 1895, months before their discovery by Roentgen. Later, Becquerél reported gross skin "burns" from radioactivity exposure in 1901, and Grubbé recognized fluoroscopic injury in 1903. The seriousness of radiation hazards, however, was incompletely understood for many years, and two decades elapsed before the establishment in Britain of the first X-ray and Radium Protection Committee in 1921. Other national groups were formed soon thereafter, and the International Commission on Radiation Protection was established in 1928. More recently, the rapid growth of medical radiology and the development of atomic energy programs have led to considerable public interest in radiation safety, and in the United States there has been legislation to control radiation on federal, state, and local levels. During the past twenty years, genetic injury to mankind as a whole, in addition to injury to the irradiated individual, has received considerable attention.

Ionizing radiation is potentially hazardous to users, patients, and other exposed persons, but any risks involved must be viewed in proper perspective. X-rays and radioactivity are important tools in industry and indispensable to modern medical practice. It is unthinkable to discontinue their use, as long as the associated risks can within reasonable limits be controlled. As we shall see, adherence to established rules reduces such risks to extremely low levels. In technologically advanced societies many other health hazards exist, such as polluted air and water, contaminated food, traffic, high levels of noise, and our propensity to overindulgence. Of all present day hazards, those from ionizing radiation have likely been most thoroughly studied, and extensive clinical experience and research span a period of over seventy years.

Two basic problems must be solved in achieving radiation safety. The first is to determine "maximum permissible levels" of exposure within which risks of injury are considered acceptable by most informed medical and other scientific authorities. The second is to

control radiation sources and their use in practice so these permissible levels are not exceeded.

This chapter considers concepts involved in solving both these problems. Chapter 14 discusses actual procedures followed to achieve low radiation exposure levels, and Chapter 15 presents illustrative calculations of radiation shielding barriers employed in x-ray and high-activity radioactivity installations.

MAXIMUM PERMISSIBLE DOSE EQUIVALENT (MPD)

Before considering specific MPD values, we shall first discuss two more basic topics: the types of radiation injury, and the rem, the special dosage unit used in radiation protection work. As a matter of considerable interest, data on actual dosage levels found in clinical practice are also included.

Types of Radiation Injury

The user as well as other persons may be exposed when ionizing radiation is used. Of course, patients receiving diagnostic and therapeutic procedures are intentionally irradiated. An important objective in such procedures is to minimize the resulting exposure to the extent consistent with obtaining the desired clinical results. When radioactive materials are used radionuclides may be released in various ways and later reappear as contamination, sometimes remote from the source. In small amounts contamination is primarily a nuisance in sensitive applications (examples are fallout damage to photographic film and high background in low-level radioactivity measurements). Greater amounts in air, water, and food supplies, however, may present potential health hazards.

Fortunately, there is available a considerable body of knowledge regarding radiation hazards, gathered through medical experience with x-rays and radium, animal experimentation, accidental exposure (uranium miners, radium dial painters, early pioneers in radiology, nuclear fission workers), and military exposure from nuclear bomb explosions and fallout.

Medical experience and accidental exposures have provided considerable data, especially where dosage levels could be documented with reasonable accuracy. Animal experiments are of great interest in elucidating basic scientific information; all animal data, however, must be extrapolated to humans with considerable care.

In Chapter 5 we discussed radiation injury, particularly in radiation therapy where *substantial* exposures are given over a period of

TABLE 1. SOME SOMATIC EFFECTS OF IONIZING RADIATION^{1,2}

A. Sensitive Tissues in Humans of All Ages

1. Blood
 - (A) Circulating cells: leukocytes are extremely radiosensitive.
 - (B) Marrow cells
 1. Quite radiosensitive; 400 r may sterilize 98% of marrow cells.
 2. Leukemia induction* may result from as little as 200 r in an adult, delivered all at once or in fractions.
2. Eye lens:* radiation cataracts may result from as little as a few hundred r with a latent period of 4 months to 29 years.
3. Skin*
 - (A) Direct reaction, for patients treated to limited areas for a few days.
 1. Less than 1,000 r leaves little permanent mark.
 2. 2,000 to 3,000 r causes tanning, permanent hair loss, skin vessel damage.
 3. More than 3,000 to 4,000 r leaves skin thinned, sensitive, and subject to infection.
 - (B) Squamous cell carcinoma
 1. May develop on very rare occasions from single large dosage.
 2. Develops more often, but still relatively rarely, with repeated smaller doses to thousands of r total. Examples: occupational exposure of early workers, acne treatments with improper dosage control. Very rarely seen nowadays.
4. Gonads:* permanent sterility may occur with 500 r or more gonad dose. If exposure were accidental, this would normally result also in severe radiation illness because extensive areas would be irradiated. There is no evidence of above average incidence of sterility in radiation workers nowadays.
5. Lungs: some uranium miners exposed for years to upwards of 1000 r from radon inhalation developed lung cancer. Latent periods averaged 17 years.
6. Bone
 - (A) Radium ingestion: some radium dial painters developed osteogenic sarcoma; 3.6 micrograms and more of radium were fixed in bone. Latent periods averaged 15 years.
 - (B) X-rays: some persons irradiated to the spine for benign conditions (to 3,000 r and more) developed cancer after 4 to 22 years.
7. Thymus: as little as 300 r to a sizeable area of the chest in youngsters was sometimes followed by thyroid carcinoma, with an average latent period of 15 years.
8. Entire body: the life span may be shortened about 1 day per r.

(continued)

TABLE 1. SOME SOMATIC EFFECTS OF IONIZING RADIATION (continued)

B. The Human Fetus*

1. Impaired development: 1,000 r generally leads to miscarriage; 50 r during the first few weeks may lead to gross birth defects.
2. Childhood malignancy: statistical studies indicate 2 r during pregnancy may result in a 40% increase in frequency over that of unexposed fetuses.

C. The Growing Fetus or Child*

Bone growth is evidently retarded by epiphyseal cartilage irradiation of several hundreds of r. This fact is of clinical interest when a child's bone must be irradiated in radiation therapy, to avoid asymmetrical bone growth (scoliosis, etc.).

* Items marked by an asterisk are of greatest interest in routine medical radiation protection.

one to 50 or so days. In radiation protection work the problem is generally somewhat different. Excepting rare accidents, our primary concern is potential injury from relatively *small* exposures delivered repetitively over a long period of time, approaching a lifetime.

Two general types of injury can result when an individual is exposed. The first, damage to the various somatic tissues, can appear during his lifetime. The second follows irradiation of *gonadal* tissue; it affects the descendants of the irradiated individual because of mutations potentially transmitted by his reproductive cells. In general, much greater dosage levels are required to produce significant somatic than genetic changes.

SOMATIC INJURY. Table 1 presents a brief partial summary of the somatic effects of ionizing radiation.^{1, 2} Somatic injury may become evident in two ways. The first is by loss of tissue function following gross cellular damage, as occurs to bone marrow following acute irradiation. The second is by induction of cancer, generally following prolonged exposure to moderate dosage levels; though delayed in appearance, one to more than 20 years, this injury may be the more serious. There appears to be a greater sensitivity in the very young to both types of injury, particularly the human embryo during the first weeks of gestation.

In adults, the major radiation hazard results from irradiation of blood-forming organs. Functioning bone marrow is especially sensitive to both direct destruction and leukemia induction. Fortunately, marrow cell repopulation can occur spontaneously even after considerable marrow cell destruction with proper medical management. The induc-

TABLE 2. DISTRIBUTION OF ACTIVE BONE MARROW IN THE ADULT HUMAN³

Body Part	Percent of Total Active Bone Marrow
Head	13
Upper limb girdle	8
Thorax: Sternum	2.5
Ribs 1-12	8
Spine: Cervical	3.5
Dorsal	14
Lumbar	11
Sacrum	14
Lower limb girdle	26
Hands and forearms, feet and ankles	Negligibly small

tion of leukemia, however, is thus far an irreversible process, so it is the most serious hazard from occupational irradiation of functioning marrow cells. In adults these cells are located primarily in the torso and head, with few in the extremities (Table 2).³

The eye lens is subject to radiation cataract from relatively low x-ray dosages. Higher LET radiation (like neutrons, protons, etc.) is even more damaging, rad for rad, than x-rays. The skin is relatively insensitive to radiation.⁴ This is good fortune from a practical point of view because the skin of a radiation worker's hands is sometimes necessarily close to the radiation source and is less easily shielded than the torso.

Contrary to earlier reports, recent studies indicate radiation workers do not age substantially faster than other people. The shortening of life span appears to be of the order of one day per rem of whole body radiation.⁵ A person exposed for 50 years to the full permissible maximum of 5 rems per year might hence lose about 250 days. (More typical levels of 1 or 2 rems per year reduce this figure to 50 or 100 days.)

The very young are more sensitive to ionizing radiation than adults, in both direct injury and induction of malignant disease. For example, gross defects may be produced when the embryo receives 50 rems during the first weeks of gestation.⁶ MacMahon has compared the incidence of malignant disease in children irradiated *in utero* to an average of 2 rem (range of 0.4 to 5 rem) with that of a control unirradiated group. There was a 40 percent increase in incidence of all kinds of childhood malignant disease in the irradiated

TABLE 3. TYPICAL LD 50/30 VALUES FOR MAMMALS—REMS OF HARD X-RAY IRRADIATION^B

Animal	Dosage in rems
Guinea pig	200 to 400
Swine	275
Dog	325
Goat	350
Monkey	500
Mouse	400 to 600
Rat	600 to 700
Hamster	700
Rabbit	800
Man	See note 2 below.

Notes:

1. LD 50/30 is the massive whole body dose that kills 50% of a large number of similar animals within 30 days of exposure.
2. Man's response is likely near that of the monkey but can only be estimated from limited accident and disaster data.

group.⁷ It is evident that scheduling of x-ray studies of potentially pregnant women should receive special consideration from physicians. It has been suggested that elective x-ray examinations be scheduled immediately after menstruation where practicable.

When young mammals are subjected to epiphyseal irradiation of the order of a few hundred rads, bone growth may be retarded, a fact also observed in children. Radiation therapists are conscious of this type of potential complication in treating various childhood malignant diseases with some inevitable irradiation of the bone, and when possible they generally avoid treatment of benign conditions in children.

In emergency or disaster situations an individual may receive substantial dosage to his entire body in a single exposure. When this happens the effect may vary greatly among individuals. As a result, one speaks of the "LD 50/30" value for a species: the massive whole body dose that kills 50 percent of a large group of similar individuals within 30 days. Table 3 presents typical LD 50/30 values for mammals exposed to hard x-ray irradiation.⁸ Whole body irradiation levels greatly exceeding 25 rems are considered a serious matter; 25 rems is sufficiently traumatic to be considered inadvisable more than once in a lifetime, even in emergency situations.

Military and disaster radiation safety procedures are a specialized subject beyond the immediate concerns of this book. Injuries under conditions of war are highly complex and potentially involve radia-

tion damage from exposure to gamma, neutron, and beta rays; thermal burns to exposed surfaces; and mechanical injury from blast and collapsing structures. Our coverage is therefore directed primarily to the control of radiation injury from operation of medical and ordinary industrial installations.

GENETIC INJURY. This term normally refers to point mutations in reproductive cells. When such cells are irradiated, three types of effects can result:

1. *Very serious chromosome damage.* Chromosomal damage can presumably involve either a particularly critically important point mutation or a group of them whose combined effect is devastating to the cell heritage. The resulting progeny has very serious defects often incompatible with survival, and the reproductive cell strain may hence die out in one or a few generations. Although constituting a disaster to the involved individuals, the long-term social consequences are not serious because the defect is not transmitted.
2. *"Beneficial mutations."* Evolution has possibly been initiated by mutations and "beneficial" characters perpetuated by natural selection; these are, however, likely rare mutations.
3. *Debilitating mutations.* The majority of point mutations is believed to produce defects harmful but not incompatible with survival to the reproductive age of the individuals involved. As a result, many such persons may pass on their defective genetic heritage, which can potentially become widespread, leading to more sickly, less virile future generations.

In view of the social seriousness of the third group, *all* unnecessary mutations are generally considered undesirable, and we attempt to minimize their production.

MUTATION FREQUENCY. One would expect the incidence of mutations to increase with dosage, and this appears to be true. Working with *Drosophila* fruit flies exposed to x-rays, Muller reported the mutation rate to be proportional to the total dose and independent of the time schedule of delivery, indicating there is no recovery from genetic mutations.^{9, 10} More recent studies¹¹ show there may be some recovery, especially when dose rates lower than in Muller's experiments and more similar to those received occupationally are used. For such irradiation it is possible injury may not occur until a threshold value of dosage is reached. Much more work must be carried out to clarify this picture, however, and authorities in setting MPD values as guidelines for the use of ionizing radiation have con-

sidered it prudent to assume no threshold or recovery effects exist for genetic injury.

The "mutation rate" referred to by geneticists is the final value after continued irradiation of the interbreeding population, generation after generation. Relatively few mutations appear in the first generation because the final mutation rate is built up exponentially over many generations.

To illustrate, consider the "mutation doubling-dose" concept, referring to the average gonad dosage to an interbreeding population which, given generation after generation, ultimately doubles the natural mutation rate. The doubling dose for man is estimated to be of the order of 50 rems. (This figure is the median of an estimated range of values; the lower limit is only one fourth this amount.) Thus, if mankind received an average of 50 rems henceforth, the genetic defective frequency at birth might rise from, for instance, 2 percent at present to an eventual 4 percent. Suppose that only 5 instead of 50 rems were received. Then something closer to 10 percent instead of 100 percent more could be expected, or an increase from 2.0 percent to 2.2 percent, a much more acceptable social burden. This example assumes mankind provides no eugenic or medical correction during all these generations. Initially, the mutation rate would be to roughly 2.02, 2.04, 2.06 percent, for the first three generations, during which time some scientific advances might arrest the increase and hopefully reduce the base figure.

The International Commission on Radiation Protection (ICRP) has set a limit of 5 rems for *non-medical* exposure to man-made ionizing radiation, with the additional recommendation that medical irradiation be reduced as far as possible "without detriment to the medical value of the procedures." The question of the genetically significant age period arises. Obviously, persons above forty conceive children only infrequently, so a given hereditary defect is less likely to be transmitted by an older than by a younger person. For this reason, exposure between conception and 30 years of age has been taken to be the genetically significant period; it is assumed half the children are born of parents 30 years old and younger.*

Man has evidently been subjected to natural radiation dosage throughout his evolution; it is reassuring to compare the natural radiation dose with the 5 rem limit proposed by the ICRP. Table 4 shows typical gonad and bone marrow estimated doses from natural sources.¹² Note that the ICRP limit approximates natural radiation levels.

* This age of demarcation might be expected to vary with social patterns, longevity, etc. The 30 year figure is typical for most countries represented in the ICRP.

TABLE 4. TYPICAL DOSAGE FROM NATURAL SOURCES:
CONCEPTION TO 30 YEARS OF AGE¹²

Source		Gonads (rem)	Bone Marrow (rem)
Cosmic rays: ionizing component	0.75	1.50	1.50
	neutron component 0.75		
Terrestrial radiation: outdoors	0.55 to 11	1.5	1.5
	indoors 0.6 to 6		
Internal radiation:	⁴⁰ K	0.6	0.5
	²²⁶ Ra	0.015	0.02
	²²⁸ Ra	0.025	0.03
	¹⁴ C	0.02	0.05
	²²² Rn (bloodstream)	0.09	0.09
Total*		4	4

* Estimates are approximate only, varying greatly with locality, housing, etc.

The Dose Equivalent Unit: The Rem

It was shown in Chapter 5 that the biological effect of radiation depends not only on the average tissue dose in rads but on several other factors as well. These include LET of the radiation, the exact rad distribution in tissue, rate of delivery and fractionation, as well as biologic and chemical variables pertaining to the individual. It is evident that the rad dosage alone is of limited value in estimating biologic injury unless additional factors are employed to accommodate the other variables.

In radiation protection work one employs the "dose equivalent" rather than rad dose. The dose equivalent is intended to provide a gauge of the likely injury to the irradiated part or organism. The unit is the rem. The dose equivalent (DE) is obtained from the rad dose D by the following relationship:

$$DE = D (QF)(DF) \dots \dots \quad (13-1)$$

QF is the "quality factor," which is a number corresponding to the LET of the radiation used.¹³ DF is the "distribution factor." It is a number which takes care of the modification in biologic effect due to non-uniform distribution of internally deposited radionuclides. Other factors may also arise corresponding to other variables, and (13-1) provides for such factors as they are determined.

For most clinical work (x-rays, gamma rays, and beta rays), QF and DF are essentially unity, and the number of rads is numerically equal to the number of rems. (The slight changes in QF of x-rays at

TABLE 5. QUALITY FACTORS—WHOLE BODY RADIATION¹³

Type of Radiation	QF
Photons and electrons	1.0
Fast neutrons* and protons, alpha rays	10
Heavy recoil nuclei	20

* For purposes of protecting the eye lens, a QF of 30 is taken for fast neutrons.

very high and very low energies are ignored in protection work, where extreme precision is unnecessary.) A QF correction becomes necessary when neutrons and other high LET radiation are employed (see Table 5).^{*14}

MPD Values

Table 6 briefly summarizes MPD values set by the NCRP.¹⁵ The limits apply to man-made radiation other than that given under medical prescription. There are essentially three groups of persons for whom specific limits are set. The first is workers in controlled radiation areas containing machines and/or radioactive materials and presumably restricted in access to persons having business there. The second group consists of those outside such areas who may be irradiated as a result of operation of the installations. Irradiation may result from stray radiation to adjacent areas or the escape of radioactive materials. The population as a whole comprises the third group.

Occupational exposure MPD limits are the highest, to permit the carrying out of necessary operations in a practical way by the relatively limited number of radiation workers. The permissible values are at least ten times lower than previous levels which were established to be well below those producing detectable somatic radiation injury. MPD figures were substantially reduced about a decade ago. This was done for *genetic* reasons only, to avoid a disproportionate contribution of mutations from radiation workers. (Table 7 indicates an illustrative apportionment envisioned by some members of the ICRU when the 5 rems genetic figure was set.)¹⁶

Persons in the second group should, in general, receive less than 1/10 that of radiation workers, and the general population should receive on the average about 1/30 as much.

* In the past, the term RBE (relative biologic effectiveness) has been used in both radiobiologic research and radiation protection work, to account for LET effects. However, RBE must involve consideration of all relevant factors for real accuracy, and the term is now restricted to research work. The QF is a more qualitative term, intended for the less precise requirements of radiation protection.

OCCUPATIONAL EXPOSURE—EXTERNAL IRRADIATION. Some organs in the body may be acceptably irradiated to higher dosage levels than others because they are either less critical to survival or less sensitive to radiation injury, or both. Hence, two sets of limits are given in Table 6. The first refers primarily to gonads and blood-forming organs and includes the head, trunk, and whole body. As previously indicated, the most essential organs from a survival point of view are included, and MPD levels are relatively low. The second set (excluding the eye lens) refers to specific organ and body extremities which are less sensitive and where more liberal MPD values are essential for practical reasons.

The first group has both cumulative total and accumulation rate values. The permissible cumulative total is given by the expression: $(N-18) 5$ rems, where N is the workers age. For example, he must not have exceeded $(30-18) \times 5$ rems = 60 rems whole body dosage by his thirtieth birthday. Five rems/year is generally taken as a design figure in planning radiation protection shielding and procedures. In addition, the 3 rem/13 weeks figure is also set. It is considered undesirable to acquire exposure at a rate in excess of this figure, even when the annual total is not excessive. When a person older than 18 has not been previously exposed there may be specific situations in which the 5 rem/year design figure is usefully exceeded on a limited scale. Under such unusual circumstances, one may permissibly be exposed to up to 12 rems in a single year, for four 3 rem/13 week periods. This is not contemplated as a frequent procedure, however, and in general practice is very rare in medical installations.

Of the vulnerable specific organs, the eye lens is subject to cataract formation, especially from neutron irradiation. The skin of the entire body can conceivably be irradiated by beta radiation in air or liquid or by very soft x-rays which could not reach significant gonad or bone marrow tissues. Under such conditions the critical organ is the skin, which may receive twice the permissible marrow dose. The hands, forearms, feet, and ankles have negligible functioning bone marrow in the adult (see Table 2), and their critical tissues therefore are considered local areas of skin. Consequently, relatively high MPD values are permitted.

OCCUPATIONAL EXPOSURE—INTERNAL IRRADIATION. Workers handling radioactive materials may inadvertently take small amounts into their bodies by ingestion, inhalation, or direct transfer through the skin. The fate of these materials depends on their chemical nature, particle size, solubility, and to a lesser extent, the individual involved. In general, however, a relatively constant level of retained radioactivity may be reached in time (the body burden), maintained often by con-

tinued steady acquisition of radioactive material. N.B.S. H.69 presents computed maximum body burdens in microcuries and maximum permissible concentrations (MPC) in $\mu\text{Ci/ml}$ of water and air, for various radionuclides.¹⁷ These calculations are made on the basis of the indicated MPD levels for the various body organ systems, as shown in Table 6. Body burden limits for long-lived bone seekers like ^{90}Sr and ^{239}Pu are computed on the basis of delivering the same rem dose to bone in a 50 year period as $0.1 \mu\text{g}$ of ^{226}Ra .

It must be stressed that a worker subject to both external and internal irradiation must not exceed the MPD values assigned to any organ, for the *combined* total dose from all sources.

OTHER PERSONS. Those outside controlled areas may receive at most 1/10 the occupational MPD levels, for normal operations of the installation, from external and internal radiation sources combined. This is intended mainly to reduce gonad exposure to levels low enough that the mutation contribution from this group is not excessive.

The population as a whole should average no more than 5 rems gonad irradiation from conception to 30 years of age. Since this same amount is permitted in *one* year for radiation workers, the general population MPD is roughly 1/30 lower than the occupational MPD for whole body exposure.

MEDICAL EXPOSURE. It is noted in Tables 6 and 7 that medical exposure is explicitly excluded from MPD figures for *all* groups, reflecting a deliberate decision which appears justified for at least two reasons. First, it is felt the physician must be the judge as to the diagnostic and therapeutic procedures his patient shall receive, in the light of his unique professional responsibility and competence. Second, the only practical ways to reduce medical radiation exposures involve both increased education of younger physicians and paramedical personnel and a multidisciplinary study of methods to reduce unnecessary radiation exposure. In recent years this problem has been attacked quite vigorously from both aspects in several countries. The complexities involved make it inadvisable to impose in any way non-medical restrictions on the physician's application of radiation to patients. The doctor, however, may still be held accountable, as any other user, for the safety of personnel, neighbors, and the general population incidental to the operation of his facility.

Current Radiation Exposure Levels of the General Population

As mentioned above, the MPD figures specifically exclude both natural and medical exposure. We have already shown typical gonad dosage values from natural sources from conception to 30 years of

TABLE 6. SUMMARY OF MAXIMUM PERMISSIBLE DOSE EQUIVALENT VALUES*¹⁵

A. Occupational Exposure

I. External irradiation

(A) To the entire body, head and trunk, active blood-forming organs, and gonads

1. Accumulation total: $5(N-18)$ rems by age N years.
2. Accumulation rate: maximum of 3 rems in any consecutive 13 weeks.†

(B) To specific organs

1. Eye lens: limited as above. QF of 30 for neutrons is to be used for this organ.
2. Skin of whole body: twice that to entire body.
3. Hands and forearms, feet and ankles: 75 rems per year, 25 rems/13 weeks.

II. Internal irradiation

(A) General principle: permissible levels are to be consistent with those for external radiation, and the *total* dose in rems to any organ from both sources is to be considered the significant quantity in protection work. Maximum body burdens and maximum permissible concentrations (MPC) in air and water are set on the basis of the following averaged annual dose values in rems.

(B) Whole body and gonads: 5 rems averaged over a year.

(C) Most body organs: 15 rems averaged over a year.

(D) Thyroid and skin: 30 rems averaged over a year.

B. Persons Outside Controlled Areas

I. Radiation levels attributable to normal operations are such that it is improbable any individual will receive more than 0.5 rem in any one year.

II. Body burdens and MPC values attributable to normal operations shall not exceed $\frac{1}{10}$ that set for radiation workers, averaged over one year.

C. General Population: no strictly enforceable procedures have been set. The objective is to limit average population gonad dose rates to 5 rems between conception and 30 years of age, *including all three groups*, but excluding medical irradiation.

D. Medical Exposures: no specific control is envisaged other than constant striving for general use of the best radiation practices.

* National Council on Radiation Protection and Measurements (NCRP).

† The ICRP in addition has recommended a limit of 1.3 rem in 3 months for women of reproductive age, with subsequent dosage to delivery after recognition of pregnancy of less than one rem.

TABLE 7. ILLUSTRATIVE APPORTIONMENT OF THE 5 REM RECOMMENDATION OF MAXIMUM GONAD DOSE (AVERAGED OVER ENTIRE POPULATION)¹⁶

Group	Rem
Occupational exposure	1.0
Special group exposure	0.5
Population at large exposure	2.0
Reserve	1.5
Total	5.0

Note: Medical procedure exposures are deliberately excluded; however, the ICRP recommends such exposure should be reduced to the extent possible ". . . without detriment to the medical value of the procedures."

age (Table 4). It is of interest to also consider the current medical contribution to genetically significant gonad irradiation.

Many complex factors make an estimate of this type extremely difficult. Unlike natural irradiation, medical irradiation is delivered to only a fraction of the population at a given time; many medically exposed persons are older or very sick and do not ordinarily reproduce as frequently after exposure, so weighting factors are required; and technical variations, particularly of field size and beam orientation, can introduce enormous differences in gonad dose for the same procedure. Dosimetry presents special difficulties. And finally, the amount of average gonad dose depends on the state of medical practice in a given country, reflecting economic level and social and medical customs.

We have taken data from a 1962 United Nations report concerning estimated annual genetically significant dosage from medical radiology. Table 8¹⁸ presents a brief summary of such data, converted to a 30³/₄ year period. Little attention should be paid to the large differences reported, which no doubt reflect primarily the great difficulty in obtaining accurate estimates. Even the largest figure, however, is no greater than the natural radiation contribution (about 4 rems), while most others in countries with substantial medical radiology activity range about 1.5 rem.

Even these low figures can probably be reduced by a factor of 2 or 3 without major technologic advances, and medical contribution to the genetic burden can be brought to a level well below 1 or 2 rems for the significant period. This is generally considered to represent a reasonable price to pay for the great health benefits of medical radiology.

TABLE 8. APPROXIMATE GENETICALLY SIGNIFICANT DOSE FROM MEDICAL RADIOLOGY IN SEVERAL COUNTRIES*¹⁸

Country	Estimated Dose from Conception to 30 yrs (rem)
Argentina	1.2
Austria	0.6
Denmark	0.85
France	2.0
Germany (Hamburg)	0.6
Italy (Rome)	1.5
Japan	1.2
Netherlands	0.2
Norway	0.3
Sweden	1.2
Switzerland	0.7
United Arab Republic (Cairo and Alexandria)	0.2
United Kingdom (except Northern Ireland)	0.6
United States	4.0

* These figures are crude estimates based on limited available data. Dose totals reflect not only exposure techniques but also the extent radiologic techniques are used in each country in medical practice.

CONCEPTS OF RADIATION HAZARD CONTROL

Most sources of ionizing radiation fall within five categories: x-ray and teletherapy machines, sealed brachytherapy sources, unsealed radioactive sources, high energy accelerators (generally for protons and deuterons), and neutron generators. The last two are essentially beyond the scope of this book but are briefly discussed in Chapter 16. The first three, however, are of great medical and industrial importance. In this section we discuss how hazards arise in the use of these three types of sources, to provide background for the consideration of safety procedures in Chapter 14. We also discuss radiation monitoring procedures and devices for the verification of radiation safety.

Types of Radiation Hazards

It is convenient to consider radiation hazards against the background of the MPD requirements in Table 6.

OCCUPATIONAL EXPOSURE—EXTERNAL IRRADIATION. Penetrating radiation endangers the blood-forming organs and gonads, particularly

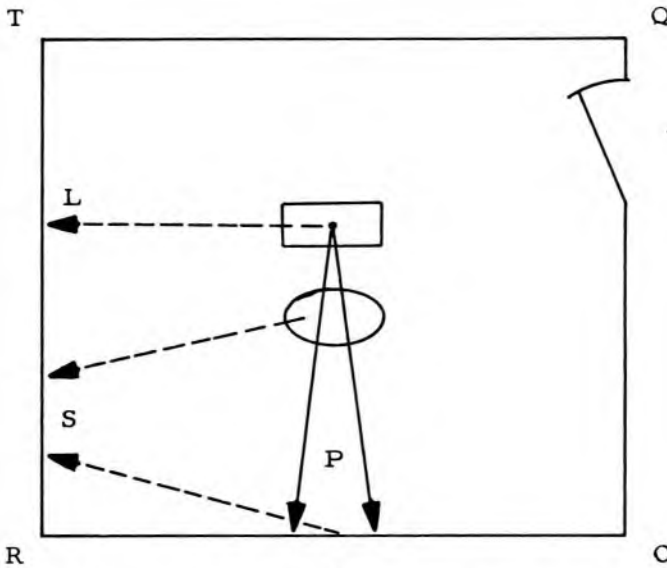


Fig. 1. Types of radiation present in an x-ray room. P—primary beam directed towards the patient. S—radiation scattered from patient and other objects struck by the primary beam P. L—leakage radiation from the tube assembly.

during irradiation of extensive areas of the torso. Such irradiation is possible from x-ray machines and gamma-emitting radionuclides.

Figure 1 shows an x-ray tube housing with the beam directed at wall OR. The useful or primary beam P strikes the patient, and the unattenuated portion reaches the wall beyond. In addition to P, scattered radiation S and any leakage radiation L are also present in the room. In most x-ray work, structural barriers are provided to protect the operator and others in and around the area. Besides the walls, doors, and windows, floors and ceilings may also require special shielding. The design of such barriers is briefly considered in Chapter 15.

In medical work patients are always in the room while x-ray machines are energized. The tube housing must therefore greatly reduce leakage radiation in all clinical applications. In fluoroscopic and some other procedures, personnel must be in the room as well. Additional special equipment and accessories are then required.

Teleisotope machines require room and housing shielding requirements similar to those of x-ray machines. In addition, the housing and shutter shielding must reduce leakage to very low levels even when the machine is in the "off" position, since radioactivity cannot be turned off like x-rays. Other radioactive sources include both brachytherapy and unsealed types. In handling such sources the basic rules apply, of maximizing distance and speed during work and using shielding when indicated. Special shielded work arrangements

are often required to store, handle, and transfer gamma emitters. Most man-made radionuclides are subject to AEC regulations (see Chap. 14).

Both hard and less penetrating radiation may endanger superficial parts of the body. Of great historical interest is occupational injury to the hands during fluoroscopy and radium therapy. Nowadays, appropriate procedures and accessories make such injury extremely improbable. Injury to the eye lens can also potentially result from any type of ionizing radiation. (High LET sources like neutron beams are especially injurious.) In medical work goggles are sometimes desirable to shield the eyes from high energy beta rays. In other work normal precautions which protect the head and torso incidentally protect the eyes as well.

OCCUPATIONAL EXPOSURE—INTERNAL IRRADIATION. Radioactive materials inside the body may constitute a very serious hazard when present in significant amounts. They may originate in both sealed and unsealed sources.

Sealed sources are relatively unlikely to contribute occupational internal irradiation, although difficulties may arise under two types of circumstances. A source and/or its housing may, during manufacture and installation, acquire surface contamination which finds its way into the body. Also, sources can be mechanically damaged and release their contents. Fortunately, the contents of most sealed sources are minimally toxic radiochemically, with the major exception of some old radium sources. (Modern sources contain insoluble radium sulphate, but some older sources contain the more soluble chloride or bromide.)

Any unsealed radioactive material can potentially enter the body by ingestion, inhalation, and transfer through the skin. Careless handling can result in direct ingestion. The fate of such materials depends on many variables.¹⁹ In general, however, the alimentary tract is always irradiated. Absorption into the blood results in specific organ irradiation (bone marrow, liver, spleen, thyroid, and generally the kidneys as well). Both dust particles and aerosols may be inhaled. When insoluble particles exceed 10 micra in diameter they tend to be swallowed rather than deposited in the lungs. Smaller insoluble particles may penetrate the lung alveoli and remain there for some time. Soluble materials tend to be absorbed into the blood either through the lungs or alimentary tract. Radioactive material is transferred across the skin primarily following injury by abrasion or actual puncture.

The evaluation of low levels of internal radioactivity is a highly specialized procedure requiring special equipment and experience.

In addition, the removal of long-lived radionuclides is extremely difficult once they have become "fixed." As in dealing with other hazards, prevention is the best approach. Immediate expert decontamination, body burden evaluation, and medical management are desirable in the unlikely event of significant contamination.

PERSONS OUTSIDE CONTROLLED AREAS. Radiation protection barrier and other design is based on meeting the 10 times lower MPD figures for people who are essentially "neighbors." This requirement increases the necessary thickness of walls or other barriers facing uncontrolled areas next to radiation installations for penetrating radiation such as that from x-ray and gamma-ray sources (by a tenth value layer). Brachytherapy and unsealed gamma-ray sources may require special care in their use and transfer because in use they are freely movable and substantially unshielded. A most distressing potential hazard of radioactivity is the spread of contamination beyond the confines of the installation.

Special problems arise when patients are treated with millicurie quantities of radionuclides. Sources may be lost or damaged; other persons, including his family, other patients, and personnel, may be irradiated by the patient receiving treatment; and surgeons or pathologists may be exposed during emergency surgery or at autopsy.

PATIENTS. The current average genetically significant radiation dose from medical procedures can probably be reduced by a factor of at least two in most countries. The basic approach is twofold: reducing unnecessary use of radiation and gonad dosage associated with *necessary* procedures.

The first objective involves both judicious ordering of diagnostic studies and their proper preparation and execution to minimize repeated studies. In addition, radiotherapists now generally avoid treating benign conditions when alternatives exist.

The reduction of gonad dosage in x-ray diagnosis is primarily a matter of field size limitation and the use of more sensitive radiation detectors. Most nuclear medical studies do not contribute substantial genetically significant gonad irradiation.

Radiation Safety Monitoring

The term "monitoring" refers here to procedures for verifying that radiation safety has been achieved. The three most commonly employed methods are visual inspection, area monitoring, and personnel monitoring.

Often, a trained observer can quickly detect improper equipment, procedures, and working arrangements by visual inspection.

For example, direct examination and simple checks may reveal mechanically damaged cones, collimators, fluoroscopic shutters, and filters; insensitive fluorescent screens; torn lead rubber aprons and gloves; etc. Bad practices and work arrangements may also be easily recognized. These include exposing personnel to the primary beam and using unnecessarily large fields in x-ray work, and employing poor storage facilities, tools, and handling procedures in radioactivity work.

To be sure of radiation safety one must ultimately conduct measurements: area and personnel monitoring. Area monitoring

TABLE 9. RADIATION SAFETY MONITORING DEVICES

A. For Area Monitoring, Primarily

1. Ionization chambers

(A) Cutie pie: portable survey meter. Sensitivity to range as low as 0-5 mR/hr possible with 500 ml volume chamber. Photon response uniform from 40 keV to 2 MeV.

(B) Fixed units: laboratory units (Lauritsen electroscope), for gamma-emitter assay. Remote detector units, for monitoring dangerous areas.

2. G-M counters: all units have energy dependence and coincidence errors.*

(A) Portable: small units are used for general laboratory monitoring. High sensitivity: 0-0.2 mR/hr is easily attained for radium gamma rays with small G-M tubes.

(B) Fixed: alarm systems in measurement areas detect the intrusion of high activity gamma-emitting sources which can increase laboratory background excessively.

3. Scintillation detectors: alpha detection units are available as well as more sensitive photon detectors. More costly than G-M units.

B. Personnel Monitoring, Primarily

1. Film badges: most generally used for routine monitoring of all types of radiation. Accuracy limited.

2. Pocket ionization chambers

(A) Both minometer and self-reading types.

(B) Used to measure x-ray and gamma-ray exposure during specific operations for training and technique improvement purposes.

(C) Relatively accurate indications, especially if energy corrections are made at lower photon energies.

3. Pocket G-M unit with "howlers": alarm unit only, photon and neutron detection mostly.

4. LiF and other detectors: currently under investigation as substitute for film badges.

* See Chapter 10.

evaluates radiation intensity and radioactivity contamination levels in and around an installation. The results establish the safety of the *location*. One is, however, primarily interested in the safety of *individuals* who work in that location; their exposure depends not only on radiation levels prevailing in the area but also on the skill with which workers carry out their duties. A careless or unskilled person may receive 50 millirems upper torso irradiation during a particular radium procedure, whereas another worker may receive only 15 millirems at the same task. Evidently, one must monitor the individual to both verify safety and properly train personnel.

We shall now discuss various monitoring instruments and procedures.

AREA MONITORING. Three types of radiation detectors, ionization chambers, G-M tubes, and scintillation detectors, are employed in area monitoring instruments.

Portable ionization chamber survey instruments are extensively used for measuring x-ray and gamma-ray intensity levels (see Table 9). For example, a cutie pie may be used to measure radiation levels near a fluoroscopic x-ray machine, around a 250 kV or supervoltage x-ray or teletherapy room, near a radium storage and handling area or a "hot" radionuclide laboratory, around a patient receiving radium therapy, etc. Such instruments give a rate indication, usually directly in mR/hr. They have the advantage of relative energy independence, from about 40 keV to 2 MeV. This results from two facts: the cylindrical chamber is made of low *Z* materials and is operated as a true ionization chamber. Considerable sensitivity (5 mR/hr full scale) is made possible by the use of a large air volume of about 500 ml. This construction presents a disadvantage in measuring beams of narrow cross section, however, yielding *low* readings which may underestimate the hazard (see Fig. 2).

G-M tubes are much more sensitive than ionization chambers of equal size, for reasons discussed in Chapter 10. They are widely used in portable survey meters for x-ray and gamma-ray detection at lower intensity levels (0.2 mR/hr full scale and below from radium gamma rays). However, their indications are basically in counts per minute (c/m) only, not dosage rate, because G-M tubes are highly energy dependent. Inexpensive portable G-M survey meters also often read grossly in error even in measurement of gamma rays used to calibrate them. (Ten units of one type have been found to read consistently high by factors of 3 to 5 times.) In spite of their limitations G-M survey meters are used to check well-shielded therapy installations whose leakage radiation exposure rates are very low. Fairly quantitative results are often obtained by a simple but judicious inter-

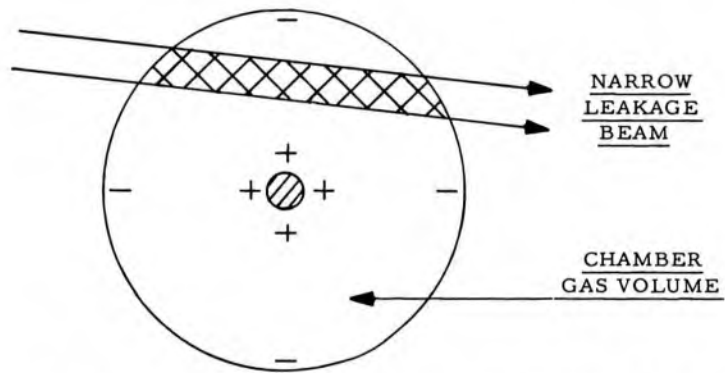
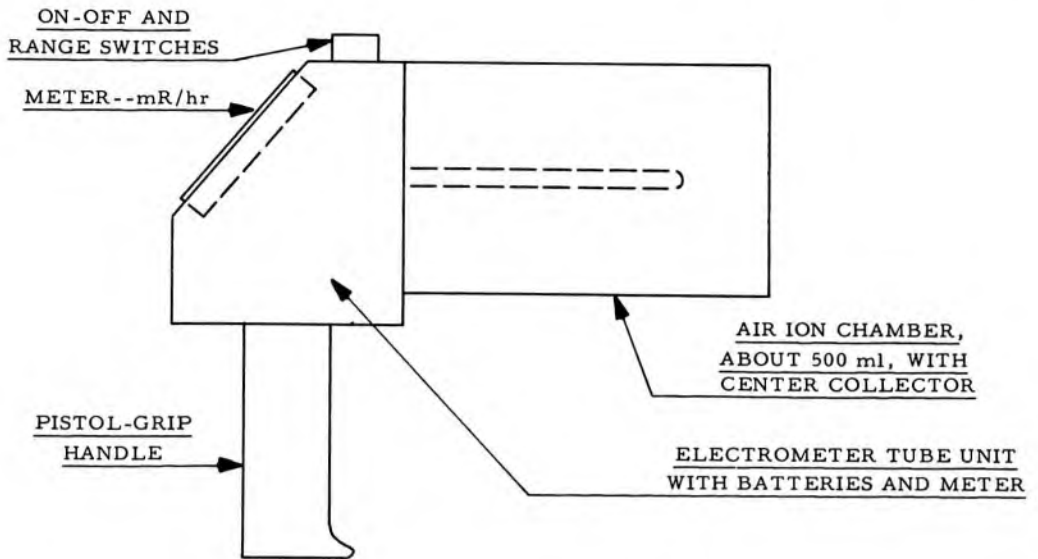


Fig. 2. Cutie pie unit. Top. This is a self-contained portable ionization chamber meter. Although only moderately sensitive (as low as 5 mR/hr full scale), it is relatively photon energy independent. Bottom. A large gas volume is required to yield useful sensitivity. This causes the instrument to read low in measuring narrow beams.

comparison of the G-M unit readings with those of a cutie pie at the same point.

Portable end window G-M units are invaluable to check contamination of laboratory surfaces and people, particularly where beta-emitting radionuclides are involved. Table tops, floors, and walls, as well as garments, bedclothing, tubing, tools, and skin—are all quickly checked with great ease and sensitivity.

Ordinary scintillation crystal detectors are limited to gamma-ray detection because their crystals are encased in metal. They can be made more sensitive to gamma rays than G-M tubes, however. Other scintillation units (screen types) have been developed specifi-

cally for alpha detection. They are used to detect ^{222}Rn and its daughters and other alpha emitters with great sensitivity.

PERSONNEL MONITORING. As previously indicated, accurate personnel monitoring provides the most convincing evidence a worker has not exceeded the MPD values from external radiation sources. (Dosage levels from internal irradiation must be inferred from whole and partial body counting and complex calculations, necessarily limiting our discussion to external sources.) Actual cumulative personnel exposure records are generally required by law for radiation workers unless proven unnecessary. Personnel monitoring is thus not only good safety practice but a medicolegal necessity.

All personnel dosimeters yield only crude indications of the dosage levels to the most critical organs, bone marrow and gonads. Since such devices are necessarily worn on the person rather than within the body, they at best read skin dose. Failla has shown this to be a rough approximation to marrow dosage for all but very soft photon radiation in clinical radiology.²⁰ It must be stressed that no one figure can give both marrow and gonad dose, and skin and marrow dosage values differ by a factor of two or more. In addition, substantial measurement errors can arise, particularly in film dosimetry. All legal interpretations of personnel monitoring records should therefore be reached only after expert testimony to evaluate the raw data.

Both film and ionization chamber monitoring devices can be used for personnel monitoring (Table 9).

FILM BADGES. Film badges, consisting of calibrated dental films mounted in special holders (Fig. 3), are employed for most routine work. Nowadays, most commercial services provide film inserts to be replaced at regular intervals, usually once or twice a month. The badge is worn at an appropriate location on the torso and the dosage inferred from densitometry of the processed film. The use of multiple small filters in the plastic film holder enables the supplier to provide a rough estimate of the total badge exposure contributed by each type of beam (i.e., β -rays, soft x-rays, hard x-rays, and γ -rays).

Film badges, unfortunately, yield only crude estimates of radiation dosage, even under the best of circumstances. Several causes contribute to this inaccuracy, of which the following are among the most important:

1. *Photon energy criticality of film*, which makes calibration at various beam qualities extremely difficult despite the use of filters in the badge.
2. *Calibration inaccuracies.*

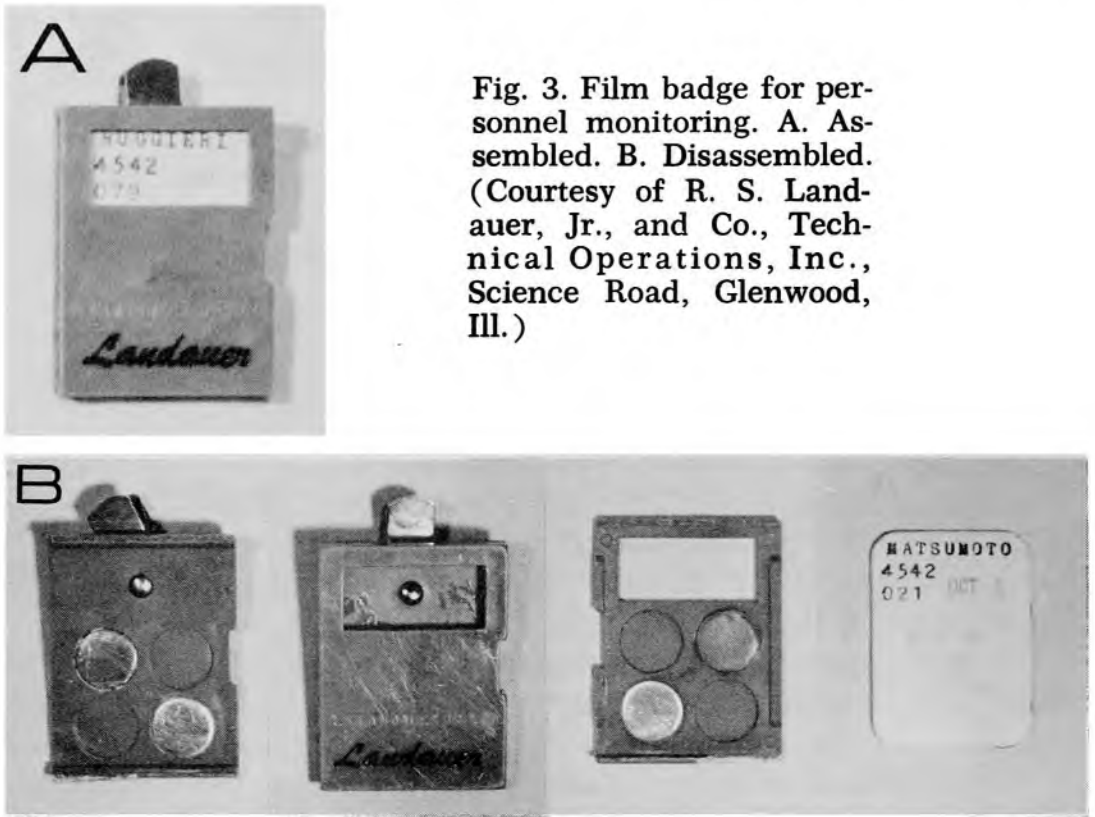


Fig. 3. Film badge for personnel monitoring. A. Assembled. B. Disassembled. (Courtesy of R. S. Landauer, Jr., and Co., Technical Operations, Inc., Science Road, Glenwood, Ill.)

3. *Film variables* of development, chemical fogging, and image fading (especially in damp and hot weather).
4. *Low readings*, from beam obliquity in traversing badge metal filters.

Studies have shown that an indication ranging between half and twice the true skin dose value is probably the best that can be consistently expected from any film method.²¹ Film badges nevertheless provide the most practical routine personnel monitoring method currently available. Film badge readings are generally accepted medicolegally, when accompanied by expert testimony, and are hence very useful in spite of their basic inaccuracy.

Both adequacy of performance and long-term financial stability of commercial film badge companies are of obvious great practical interest.

OTHER PERSONNEL DOSIMETERS. The shortcomings of film badges have led to a search for more accurate and consistent radiation detectors, a difficult development because of the many practical and scientific problems involved. The limitations of film badges are serious, however, and most likely a substitute system for routine personnel monitoring will be developed in the next several years.

Any film badge or other *cumulative* dosimeter unit suffers from a basic limitation—it cannot warn the worker of dangerous exposure

until possibly weeks after the event has occurred. Pocket ionization chamber instruments offer a faster indication and are therefore employed to quickly evaluate specific hazards from photon radiation received within a film badge period. They are particularly useful for studying hazardous procedures, such as radium or other radionuclide therapy work, fluoroscopy, etc. They give a reading in mR total dose and are fairly energy independent.

Pocket chambers are about the size of a fountain pen, with clips to hold them in place (Fig. 4). They are essentially cylindrical ionization chambers made of aluminum alloy.

Such devices have some capacitance, so they are really condenser ionization chambers of a sort. (There are two electrodes separated by an insulator; some units also contain an attached capacitor.) They are charged to a reference voltage, worn during the period of interest, and the resulting loss of voltage read to determine the received dose. The most commonly used range is 0-200 mR in clinical work.

The non-reading, or "minometer," and the self-reading are the two types of available units (see Figure 4). The non-reading unit consists of a condenser chamber only and is usually referred to as a "pocket chamber." It is charged and read by means of an associated instrument called a minometer), like the familiar thimble chamber dosimeter. The self-reading unit requires a separate charger. It can be read directly using a built-in string electroscope and microscope, by pointing the chamber at a bright background. It is referred to as a "pocket dosimeter."

The non-reading chamber unit is considerably less expensive but requires a separate reading instrument, so the initial costs of complete instrument sets are comparable. The simple ion chamber unit, however, is much cheaper to replace if it is damaged. The self-reading unit is much more convenient in that the doses received during individual procedures can be separately measured without recharging. (The simple unit must be recharged after each reading for best accuracy.)

Recently portable G-M units have been developed for personnel monitoring. These ingenious devices are only slightly larger than a fountain pen. Instead of a numerical indication they give a sound of varying pitch depending on the radiation intensity. They provide an automatic, immediate warning of radiation danger in a dramatic way not likely to be overlooked. Battery operation for as long as six months is routinely observed.

Special modifications of film badges and monitor ionization chambers or G-M units exist for mixed and neutron beams. Such devices require special competence to interpret readings and are beyond the scope of this book.

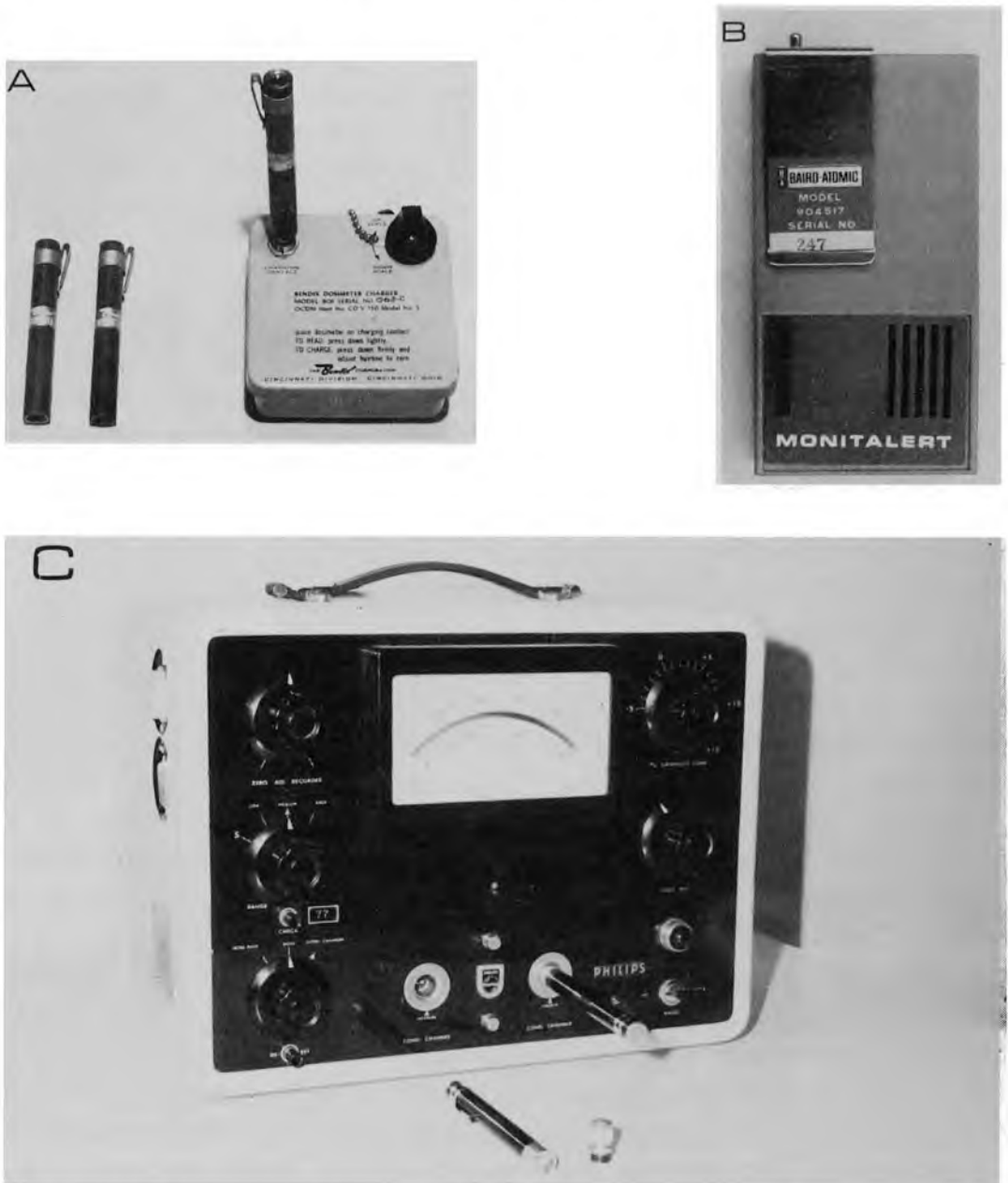


Fig. 4. Other personnel monitoring devices. A. Pocket dosimeters and charger unit.²² These devices contain their own built-in electroscopes, microscopes, and dosage scale. The unit is read while aimed at a bright light source. B. Portable G-M pocket device which produces an audible warning signal.²³ C. Reader-charger, with two pocket ionization chambers shown, one inserted into unit.²⁴

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14

Minimizing Exposure to Ionizing Radiation

In the last chapter we dealt with general principles of radiation safety. This chapter presents specific methods for controlling x-ray and radioactivity exposure.* Although the main emphasis is on medical uses, we also consider some industrial and laboratory applications. We shall, however, confine ourselves to problems originating with photon and radioactive sources; neutron and very high energy particle sources are considered beyond this book, although a few general comments are included in Chapter 16.^{2, 3}

Hazards from a radiation source vary during its lifetime. It is therefore convenient to consider safety procedures involved in the following five stages: source preparation and installation, initial check-out, use, periodic checks, and disposal. The first four of these stages are equally relevant to x-ray and radioactive sources. Disposal of radioactive sources, however, is generally more troublesome than discarding old x-ray machines. Radioactivity cannot be turned off like x-ray units, which are easily de-energized and dismantled. As we shall see below, strict rules are necessary regarding radionuclide disposal.

This chapter contains sections on x-ray machines, sealed and unsealed radioactive sources, and emergency procedures, in that order. A fifth section deals with important medicolegal administrative requirements.

X-RAY MACHINES

There are certain stages at which special attention is required to control radiation exposures from x-ray installations.⁴ A good start is achieved by selecting a well-designed machine and accessories and,

* Extensive specific references will be made in this chapter to several NCRP reports which provide the scientific basis for AEC and state codes, to which we shall also refer. The text by Braestrup and Wyckoff provides a general coverage.¹

in almost all permanent installations, providing adequate structural protective barriers. Sometimes one encounters defective or improperly installed machines and barriers; "checkout" is therefore needed upon completion of the installation, to verify both proper equipment function and effectiveness of radiation shielding. Some measurements such as of beam intensity are required as well. Proper procedures are required in the use of all x-ray machines to protect personnel, and control of patient exposure is also vital in medical work. Finally, periodic checks may be necessary to assure proper procedures, equipment operation, and, in some cases, shielding. In any of these steps the services of a qualified expert may be advisable. (See Appendix C for definitions of radiation protection terms.)

We shall consider several types of applications in the following order: x-ray therapy, x-ray diagnosis in general, medical radiography, medical fluoroscopy, and other applications.⁴

X-ray Therapy

X-rays are available with photon energies much less than 50 keV and greater than 24 MeV. There is, consequently, a wide variation in details of safety designs, but we may apply certain basic principles to almost all therapy applications. Table 1 presents a summary of the essentials. Some items of this table require further explanation and are considered specifically below.

PREPARATION AND INSTALLATION. Secondary as well as primary structural barriers are almost always required to shield x-ray therapy machines. Unfortunately, one sometimes still encounters installations with inadequately shielded ceilings and windows. Scatter is the main secondary radiation problem in conventional therapy, since leakage is readily controlled below 500 kV; however, supervoltage tube housings have substantial leakage which is very penetrating. The required thicknesses of supervoltage secondary barriers are determined therefore by leakage rather than scatter considerations. (Recall from Chapter 4 that scatter is greatly modified for such beams.) For example, a 12-inch thick concrete wall effectively attenuates scatter from a particular 6 MeV machine, but a 24-inch wall is needed to adequately attenuate its leakage radiation.

Fixed structural barriers are preferred to movable lead screens because the latter can be shifted inadvertently out of the beam or even removed altogether. At higher photon energies movable screens are not recommended at all. The operator is, in general, best protected from scatter by locating the control area outside the treatment room altogether.

TABLE 1. RADIATION PROTECTION RECOMMENDATIONS
IN X-RAY THERAPY

A. Preparation and installation*

1. Barriers

- (A) Both primary and secondary shields are generally needed.
- (B) Movable lead screens are unsuitable above 125 kV, not recommended even for lower kilovoltage operation.
- (C) Control station for 150 kV and above must be either in a completely shielded booth or outside the treatment room.

2. Interlocks

- (A) They are required for machines operable at 150 kVp and above.
- (B) They must reduce intensities to prescribed levels when door opens. Switches to restore x-rays must be on control panel.

3. Communication with patient

- (A) It must be possible to view the patient and machine controls from same position.
- (B) Verbal communication is desirable.

4. Beam direction limitation

Mechanical stops and/or electrical switches are required on x-ray units operable 150 pkV and above to limit the beam to primary barriers.

B. Checkout, generally by QE with written report

1. Installation as a whole

- (A) Area radiation survey to measure exposure rates in relevant locations.
- (B) Check operation of interlocks, filter, shutter, and other switches and signals.
- (C) Check collimating devices as to their mechanical operation and attenuation adequacy.
- (D) Patient communication and other miscellaneous items.

2. Calibrations of machine output and HVL

3. Equipment and Accessories

(A) X-ray field control

- 1. Tube housing: therapeutic type, rigidly positioned during therapy
- 2. Tube mounting: tube must not slide or rotate relative to housing window.
- 3. Collimation: permanent cones and diaphragms require same shielding as housing; however, adjustable and removable types may transmit up to 5% of the useful beam intensity for maximum HVL employed.
- 4. Light beam field indicating system: x-ray and optical fields must align properly.

* Machine specifications and tests are indicated under *Checkout*.

TABLE 1. RADIATION PROTECTION RECOMMENDATIONS
IN X-RAY THERAPY (continued)

(B) Exposure control

1. Automatic timer, or exposure or absorbed dose meter must terminate exposure at the preset time or dose.
2. For outputs of 1,000 rads/min or greater at the shortest TSD, a timer with as little as one second error is required.
3. Shutters must be used in machines operating below 500 kV if more than five seconds time is required to reach full intensity. Shutter position indicator is required on front panel.
4. Beam output monitor is required unless use is precluded by filtration requirements.
5. For tube current panel indication, mA meter or similar indicator is required.
6. Locks are required on all x-ray machines to prevent unauthorized use.

(C) Filter system

1. It must be designed to minimize the chance of error in selection and alignment of filters.
2. Removable filters must be marked with thicknesses and materials.
3. Filter slot leakage should be no more than 1 R/hr at one meter. If slot leakage beam may reach patient, no more than 30 R/hr at 5 cm from housing is permissible.

C. Operation

1. It must be in compliance with the report of the QE (qualified expert).
2. In general, only the patient may be in the room with the x-rays on.
3. *When patient must be held*, use mechanical devices whenever feasible. If patient must be held by someone, that person must be adequately shielded with all parts of his body out of the beam and as far from it as possible. His exposure must be monitored.
4. Both patient and control panel shall be observed during treatment.
5. Operating personnel must be monitored as to radiation exposure.

D. Periodic Checks

1. Calibration of beam intensity by QE is required
 - (A) At least annually unless spot checks indicate adequate constancy.
 - (B) After x-ray tube replacement or moving of machine.
 - (C) After major mechanical or electrical alterations of the tube, its housing, power supply, or controls.
2. Spot checks or measurement at a single set of factors
 - (A) Periodically, at least once a month or every 50 hours of operation.
 - (B) A permanent log must be kept.
 - (C) The instrument employed and technique must be of acceptable accuracy.
3. Regular check of mechanical condition of machine and its accessories.

The patient must be under observation during treatment. Special windows, mirrors, or closed circuit television systems are employed to permit viewing without radiation exposure. Most windows are made of high lead content glass of appropriate thickness. Transparent solutions of lead salts have also been employed. When maze shielding barrier arrangements are employed, mirror systems are sometimes useful. Closed circuit television systems offer the advantage that several receivers may operate from a single pickup, permitting supervisors and therapists as well as the machine operator to observe patients during treatment as well as treatment planning and setup operations.

CHECKOUT BEFORE USE—GENERALLY. Before routine use every x-ray therapy installation should be checked carefully. This procedure is usually carried out by a qualified expert (QE). (See glossary of terms—Appendix C.) Both the over-all installation and x-ray equipment are evaluated and a report submitted in writing with recommendations for changes as required.

The “installation as a whole” evaluation is often called a “radiation protection survey.” At the heart of this survey is a systematic measurement of exposure rates to relevant locations in and around the therapy room. This must establish that MPD levels are not likely to be exceeded for the conditions of machine operation: maximum kV and mA, operating time, and all possible orientations of the tube housing. (Mechanical stops or electric switch devices may be required to prevent directing the primary beam at secondary barriers.) In the course of conducting this survey the various switches and beam angulation limits are visually checked, as well as operation of the collimator and patient communication system.

Before use, therapy machines must be calibrated as to output rate and beam quality. Output rate may be expressed as either exposure rate (roentgens per minute) or absorbed dose rate in a phantom at a reference depth (rads per minute). Figures must be provided for each combination of beam quality, mA, SSD or SAD, field size, and collimator system. (Absorbed dose rate is preferred for units operated at 4 MeV and above, for which the roentgen is not a very useful unit.) The QE and therapist must be sure that factors used in the calibration accurately reproduce those used in actual therapy.

During the survey and calibration the qualified expert also checks certain safety aspects of the x-ray machine and its accessories relating to the control of the x-ray field, exposure, filters, and beam orientations.

CHECKOUT OF X-RAY FIELD CONTROL. A therapeutic type housing is required to limit whole body exposure to the patient. For conventional

therapy machines (below 500 kV), this means less than one R/hr leakage radiation is delivered, in any direction, one meter from the target, with the tube operating at maximum kV and mA. This limit becomes increasingly difficult to meet as operating kV is increased, for two reasons. Shielding is inherently more difficult with harder radiation, and the output rate at higher MV values is very great, with typical values for 6 MV ranging from 150 to 300 rads per minute at equilibrium depth in the patient. Actually, one is primarily interested in keeping patient dosage outside the treatment field small relative to that in the central ray. For this reason, leakage from a therapeutic housing in a machine operated 500 kV and above must not exceed one roentgen or 0.87 rads in air per hour, *or* 0.1 percent of the useful beam dose rate at one meter from the source, *whichever is greater*. For example, in a 6 MV linear accelerator with 200 rads per minute output in air at a meter, the permissible 0.1 percent would be 0.2 rads/min or 12 rads/hr, almost 14 times the 0.87 rad/hr figure! (This is perfectly reasonable, for two reasons. The 6 MV beam output is 3 or 4 times greater than that of orthovoltage beams and the percent depth dose is roughly 75 percent greater at a 10 cm depth. Hence, for a given tumor dose in rads, the 6 MV leakage dosage is only $14 / [(1.75) (3.5)]$ or about twice as great. Lateral scatter, however, is lower by an even greater factor, so a net reduction in whole body dose results.)

It would certainly appear reasonable that the tube housing position and angulation must be reproducible, to assure delivery of the prescribed treatment. Clamps and locks wear out in use and require regular service. Rotational therapy devices must be checked for mechanical reproducibility, and one cannot overemphasize the need for regular visual checks of the machine and for portal radiographs. In most modern therapy units x-ray tubes cannot slide or rotate relative to the housing window, although it can happen in some older machines. In a major teaching hospital, for example, a particular tube was once found axially rotated 180° from its proper position. As a result "treatments" were administered with virtually no x-ray output for several weeks until the machine was calibrated. This incident illustrates one reason why therapy machines should be calibrated immediately after major repairs.

The NCRP considers *permanent* cones or diaphragms essentially extensions of the tube housing; they must provide the same shielding as the therapeutic housing. Adjustable diaphragms and removable cones are permitted a much greater transmission: 5 percent of the central ray intensity is allowable, for the hardest beam employed. This is a concession to making removable units of reasonable size and weight. It should be stressed that some orthovoltage machine cones

and adjustable collimators may become bent and otherwise misaligned during shipment, installation, and use and therefore must be checked frequently.

On many adjustable collimators an optical light beam system indicating the x-ray field size is provided. Such systems are very convenient for setting up treatment portals, but provide only an approximation to the x-ray field. There are several reasons for this. The optical system is subject to variations with bulb replacement, mirror shifts, and other changes. There are penumbras in both the x-ray and light beams, whose sizes do not necessarily correspond. Finally, there can be significant misalignment of the light field with the true x-ray field. In practice, one checks the basic operation of the optical system and then the alignment. The former procedure is simple but varies with the particular unit. The second is usually verified by exposing with minimum scatter a suitable speed x-ray film (using adequate plastic absorber to assure electron equilibrium of supervoltage beams). The edges of the light field are easily identified by lead letters. The author uses letters N, E, W, and S to indicate the four directions of a rectangular field. Densitometric measurements of the developed film provide a measurement of any shifts present. Misalignment of several millimeters is not atypical, so this measurement is strongly recommended both before commencing routine work and periodically thereafter.

CHECKOUT OF EXPOSURE CONTROL. The exposure from orthovoltage and superficial therapy machines is generally controlled by an automatic timer. This is normally set and the x-rays turned on; the timer then terminates the exposure after the preset time. With very high outputs such as in some superficial therapy applications the timer accuracy may become critical. Consider, for example, a 400 R/m output rate, when 200 R in air is desired. The required time is only $\frac{1}{2}$ minute, or 30 seconds; an error of 2 seconds produces an exposure error of $6\frac{2}{3}$ percent. A high quality timer is desirable for *all* conventional therapy; the actual x-ray exposure time should be checked periodically using a good stopwatch and certainly when the machine is calibrated. (Timer switch clutch mechanisms may slip, if defective, and unduly prolong treatment exposure times.)

Orthovoltage machines often have a substantial "voltage buildup time," during which the tube voltage increases gradually to maximum value. If a shutter is not used this can result in a significant timing error in machines with high output rates. A fast shutter is therefore often used to solve this problem. The timer mechanism is generally activated when the shutter opens and the exposure terminated as above after the preset time. A shutter position indicator minimizes the

chance of accidentally irradiating the patient, should something go wrong with the control circuit.

Monitor ionization chamber units are very desirable in orthovoltage units, where substantial output changes are frequently encountered. Superficial therapy machines normally operate at very low filtration, however, so it is impractical to incorporate monitor chambers. There is, fortunately, less need to monitor these machines, since superficial units generally are very constant in output if not abused.

Supervoltage machines generally employ monitor units for beam intensity or total dose evaluation. These are located beyond any beam compensation filters, if present. In practice, such units are calibrated regularly against suitable secondary standard instruments and dosage figures derived routinely from the monitor instrument reading. The importance of keeping this instrument in good repair and calibration is obvious.

CHECKOUT OF FILTER SYSTEM. The use of an incorrect or damaged filter can introduce considerable error in both dosage and quality. Usually, orthovoltage machines provide electrical switches so x-rays cannot be energized unless some filter is in place; also, a panel light indicates which particular filter is present. In some machines filters are manually removed and inserted; these must obviously be conspicuously identified. Filters may be easily damaged, since they are made of thin sheets of soft metals, and should hence be examined often.

OPERATION. The QE often specifies some machine operation limitations which must be borne in mind. One of the most important is beam orientation, referred to above.

In general, no radiation personnel may be in the treatment room while x-rays are turned on. Radiation levels may be quite high around orthovoltage and supervoltage machines; a therapeutic head leaks up to 1 R at a distance of 1 meter in an hour's time, and even higher figures may exist around supervoltage units. In cases when a patient must be held by someone during treatment, the precautions indicated in Table 1(C) must be followed and personnel exposures measured and recorded.

PERIODIC CHECKS. The calibration of therapy x-ray machines must be performed after major machine changes as well as on a fairly regular basis to assure accurate delivery of the prescribed radiation dosage. Regular spot checking and the use of monitor chambers, if carried out correctly, can help greatly in minimizing dosage errors from machine output changes.

To avoid failures of the machine and its accessories, mechanical and electrical operations should be checked routinely by the persons operating the machine. It is by no means uncommon to find heavy orthovoltage cones hanging dangerously by a few screw threads as a result of neglect of this type of simple followup.

X-ray Diagnosis in General

X-ray diagnosis, unlike therapy, is almost always performed below 150 pkV. Structural shielding is hence simpler and cheaper than in most therapy installations. Control of patient gonadal irradiation becomes possibly even more important, however, since a more genetically significant population is involved (Chap. 13).

Several protection problems are common to all diagnostic installations. We consider these first and specific radiographic and fluoroscopic problems later (Table 2).

EQUIPMENT. As in therapy units, x-ray diagnostic machines must provide good control of the x-ray field, exposure, and filtration.

“Diagnostic” tube housings are required. By definition these permit less than 0.1 R/hr leakage, measured one meter from the target for the highest operating kV and mA values. This limit is ten times lower than that for orthovoltage therapy units, apparently for two reasons. A more genetically significant population is involved, and, because low kV is used, diagnostic x-ray beams can be more effectively shielded without adding unduly to tube housing weight and bulk. Similar considerations are involved in lower permissible leakage radiation limits set for diagnostic cones and collimators.

Precise control of exposure factors helps assure good roentgenographic quality with a minimum of repeat studies. Quality exposure timers and precise kV and mA controls and meters are essential to achieve this objective. Manual exposure control permits quick termination of x-ray exposure should the timer fail. In all manual or foot controls, “dead-man” switches are required, automatically terminating exposures when the operator releases pressure.

As previously shown (Chap. 4 and 8), a reasonably great TSD is essential in all diagnostic studies to minimize skin dose on the tube side of the patient. Twelve to 15 inches is generally required for all but dental units.

As shown in Chapter 4, the required hardness and hence filtration of diagnostic beams increases with thickness and density of the body part to be penetrated. Consequently, beryllium window tubes must include enough added filtration in the beam to meet requirements in Table 2, to protect patients' skin. When any tube is used for low kV as well as ordinary studies, there is a potential danger necessary filters

TABLE 2. EQUIPMENT AND USAGE RECOMMENDATIONS FOR ALL EXCEPT DENTAL DIAGNOSTIC X-RAY INSTALLATIONS

A. Equipment

1. X-ray field control
 - (A) Tube housing—diagnostic type: less than 0.1 R/hr leakage at 1 meter from the target.
 - (B) Collimation—*all* cones and collimators must provide the same attenuation as the tube housing.
2. Exposure control
 - (A) Quality timers, with manual control for emergency exposure termination.
 - (B) "Dead man" type exposure switches.
 - (C) mA or similar meter indicator of tube energization.
 - (D) Dials or other indication of operating factors.
 - (E) TSD values at least 12 inches, preferably greater than 15 inches.
3. Filtration
 - (A) As a general rule, at least following total filtration equivalent in beam:

50 pkV and below	0.5 mm Al
50 to 70 pkV	1.5 mm Al
70 pkV and above	2.5 mm Al
 - (B) Beryllium window tubes involve special hazards.
 1. Should not be used on multipurpose machines.
 2. At least 0.5 mm Al total filtration is required permanently in the beam.

B. Guidelines for the user

1. In general only the patient should be in the room, unless an attendant is necessary for the procedure.
2. If attendant must be in the room he should be protected with proper lead rubber garments or shields, and be as far as possible from the x-ray tube, useful beam, and patient.
3. Field size, mA, and exposure times should be minimized to extent consistent with the purpose of the examination.
4. Monitor personnel.

will not be replaced during higher kV studies. Such an error can result in especially great skin dosage from a tube with a beryllium window.

It is important to check the operation of diagnostic equipment and accessories at regular intervals as well as immediately after installation. Visual examination detects many important defects. Fluorescent screens and film can be used to verify fluorescent screen sensitivity, x-ray field size, and cone or shutter leakage.

USAGE GUIDELINES. As in x-ray therapy, no one should be with the patient during diagnostic x-ray examinations without compelling reason. When an attendant must be with the patient (as during

TABLE 3. RADIOGRAPHY: EQUIPMENT AND USAGE
GUIDELINES NOT COVERED IN TABLE 2

A. Equipment

1. Beam size and position
 - (A) Cones and collimators must be calibrated for a particular TFD. Field size at 72" TFD must be accurate to ± 1 inch.
 - (B) Multipurpose machines should have adjustable collimators with accurately aligned visual indication of x-ray field.
2. Switch

The machine x-rays must not be conveniently energized with operator outside the shielded area (except for spot filming during fluoroscopy).
3. Table top and bucky tray

In combination these must not exceed 1 mm Al equivalent.
4. Beam HVL
 - (A) Most equipment should have at least 2 mm Al filter securely fixed in housing.
 - (B) Measure the beam HVL if:
 1. Inherent filtration is unknown.
 2. Filter is inaccessible for examination.
 - (C) Measured HVL must be at least as follows:

Operation kV	Minimum HVL
50	0.6
70	1.6
90	2.6

B. Guidelines for the user

1. Patient exposure
 - (A) Limit, collimate, and align the beam skillfully.
 - (B) Shield gonads where appropriate but never as a substitute for careful beam direction.
2. Filtration adequacy
 - (A) In multipurpose machines, be sure to replace filters removed for mammography or similar studies.
 - (B) Take special care using beryllium window tubes.
3. Persons in x-ray room
 - (A) Holding of patients
 1. Use mechanical devices whenever feasible.
 2. If individual must hold patient
 - (a) Protect him with lead gloves and apron, etc.
 - (b) Position him so no body part is hit by useful beam and he is as remote from beam as possible.
 - (B) During radiography, operator must be behind shield to avoid direct scatter.

fluoroscopy and in some radiographic procedures when an uncooperative or very ill patient must be held), he must be properly protected and his radiation exposure monitored.

To spare the patient unnecessary irradiation, x-ray beam field size and exposure both should be minimized to the extent consistent with a useful study. This requires proper training of personnel and the use of sensitive fluorescent screens, films, and cassettes. Image intensifiers can also reduce patient x-ray exposure to some extent when properly employed.

Medical Radiography

Most radiographic machines are installed permanently and require structural protective barriers. The useful beam is aimed downward most of the time but may also be directed laterally towards at least one wall. Patients are usually positioned on a table of moderate height or against a vertical chest or head cassette unit; the beam, therefore, can normally reach a height above 6 feet only by accident or through neglect. For this reason, ceilings and walls above a 7-foot height are usually considered secondary barriers only.

We shall summarize radiation safety aspects of equipment and usage for all radiographic equipment (Table 3), and then briefly discuss special problems arising with particular kinds of machines.

EQUIPMENT AND USAGE GUIDELINES. Cones and collimators must be calibrated for a particular TFD. As indicated in Figure 1, the field size must be quite accurately specified. Meeting this requirement requires regular field checking to be sure cones and collimators are not misaligned in use. When a machine is used for different studies, cones may be used at greater TFD values than specified, resulting in exces-

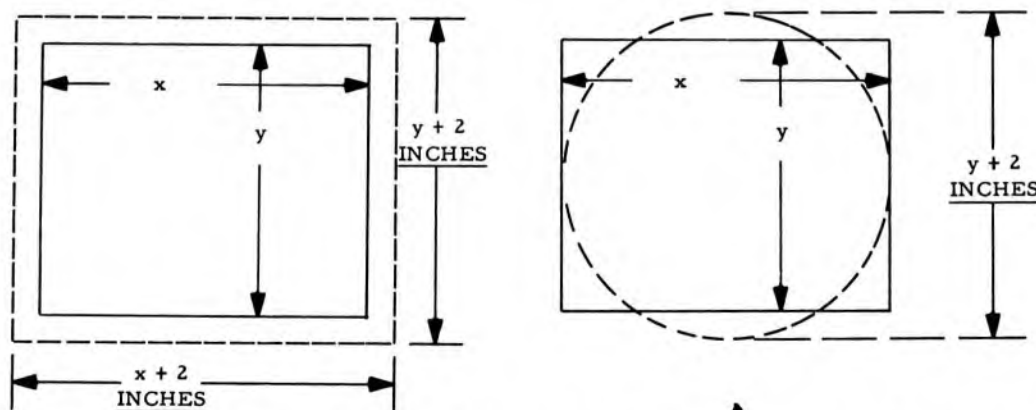


Fig. 1. Maximum permitted field size (dashed lines), for film x inches long by y inches wide; 2 inches combined overlap is allowed. Left. Rectangular field. Right. Circular field. (After NCRP Report 33⁴.)

sive field size. (See Figure 7 in Chapter 8.) For this reason, NCRP Report No. 33 recommends the use of adjustable collimators with an accurate visual indication of x-ray field size.

The beam HVL must be specifically measured unless the total filtration is known; minimum permissible HVL values are given in Table 3. For general use a 2 or 3 mmAl filter should be mounted permanently in the beam.

The second part of Table 3 presents specific suggestions for reducing patient and personnel exposure from radiographic procedures, for *any* radiographic application.

SPECIFIC RADIOGRAPHIC MACHINES. When image intensifiers are employed in ciné-radiography patient dosage may be substantial. This is especially true in units with automatic feedback controls, in which the operator may not be aware that relatively great mA or kV values are being used in examining a thick part or heavy patient. In all machines used for ciné work, skin dosage rates should be calibrated using a phantom to simulate patient absorption.

Photofluorographic machine beams are aimed at a limited fluoroscopic screen area and normally require only secondary radiation protection barriers. Three rather obvious requirements must not be overlooked, however. The x-ray beam must not overlap the fluorescent screen and its shield. Personnel generally remain in the room during exposures, since these machines are employed for mass survey work; they must therefore be protected from excessive scattered radiation. Proper beam collimation is required to avoid patient gonad irradiation by the useful beam.

Dental x-ray machines operate at relatively low kV and mA (generally 75 pkV and below, and 10 mA), reducing shielding problems somewhat, but busy installations generally require permanent barriers nevertheless. Hazards can result if one ignores the requirements of Table 4. The three equipment requirements are consistent with good radiographic practice but are often not met in older machines. In the past, injury has resulted to dentists' hands from chronic irradiation because films were held during radiography.

Portable equipment is often required when patients for various reasons cannot be moved to the x-ray department for necessary studies (see Table 5). The use of such machines creates unique hazards because structural shielding cannot be provided in the various bedside locations. As a consequence, the operator must remain remote from the patient, tube, and useful beam during x-ray exposures and should wear a lead apron. Proper training of operators is essential to protect the patient and nearby persons. The use of adjustable collimators is recommended to encourage field limitation.

TABLE 4. RECOMMENDATIONS FOR DENTAL X-RAY INSTALLATIONS

A. Equipment

1. Collimation: maximum field diameter of 3 inches at cone tip.
2. TSD: at least 7 inches for operation above 50 pkV and at least 4 inches for operation 50 pkV and below.
3. Exposure switch cord: where low machine usage permits the operator to be in the room with the patient, the switch cord must be long enough to permit the operator to be at least 6 feet from the patient, tube, and useful beam.

B. Guidelines for the user

1. Personnel safety during exposure
 - (A) Never hold film.
 - (B) Remain at least 6 feet from the patient, tube, and useful beam.
 - (C) Avoid useful beam.
 - (D) Do not touch housing or cone.
2. Patient Safety
 - (A) Avoid dental fluoroscopy.
 - (B) Use minimum patient dosage consistent with diagnosis. Use of fast film with adequate development is helpful in this regard.

TABLE 5. SPECIFIC RECOMMENDATIONS FOR PORTABLE X-RAY MACHINES

A. Equipment

1. TSD: cones or spacer frames must provide 12 inches minimum (preferably 15" or more).
2. Switch arrangement: switch cord must be long enough to permit the operator to be at least 6 feet from the patient, tube, and useful beam.
3. Where explosive gases are present, Underwriter's Laboratories approval is required for each area used.
4. If used routinely in one location only, permanent protection barriers may be required.

B. Guidelines for the user

1. The operator must stand as far from the patient, tube, and useful beam as possible; he should wear a protective lead apron or stand behind a suitable shield.
2. Portable use should be restricted as much as possible.
3. Operators should be made to understand the limitations and the proper uses of portable equipment to avoid needless exposure of patients and nearby persons.
4. No fluoroscopy is to be performed without image intensification.

TABLE 6. FLUOROSCOPY: SPECIFIC RECOMMENDATIONS REGARDING EQUIPMENT AND USAGE

A. Equipment

1. The x-ray beam
 - (A) Collimator and its cone must have same attenuation as tube housing.
 - (B) TSD: at least 12, preferably 15 inches.
 - (C) Intensity, at 80 pkV, shall be no more than 3.2 R/mA-min; preferably no more than 2.1 R/mA-min.
 - (D) HVL: at least 2.4 mm Al at 80 pkV.
2. Primary x-ray beam shield
 - (A) Attenuation of lead glass: with screen 14" from panel, and using highest kV, there must be no more than 30 mR/hr reaching 2" beyond shield per R/m at the table top.
 - (B) Beam limitation
 - (1) The shield and tube housing must be mechanically linked.
 - (2) X-rays must be turned off automatically when the shield is moved out of the beam.
 - (3) With the shield 14" from the panel or in its permanent position, and the collimator open wide, there must be a full dark margin on the screen. Similarly, the beam must not overlap image intensifier tube screens.
3. Scattered radiation
 - (A) Below the table, $\frac{1}{4}$ mm lead equivalent shielding should be provided around the table and in the bucky slot.
 - (B) Above the table, $\frac{1}{4}$ mm lead equivalent rubber flaps are recommended about the fluoroscopic screen or image intensifier tube.
4. A cumulative timer device is required.
5. Mobile fluoroscopes must always use image intensification and provide a minimum TFD of 12 inches.

B. Guidelines for the user

1. Fluoroscopy must be performed only by or under the direct supervision of a physician properly trained in fluoroscopic procedures and their hazards.
2. Patient protection
 - (A) Exposure intensity
 - (1) Should normally be 10 R/m or less, measured in air at the panel.*
 - (2) Should be periodically checked, especially in image intensifier units with automatic intensity control; in such units a suitable phantom must be in the beam during measurements.
 - (B) Unless image intensification is used, dark adaptation is always required; also, the room must be darkened.

TABLE 6. FLUOROSCOPY: SPECIFIC RECOMMENDATIONS REGARDING EQUIPMENT AND USAGE (continued)

- (C) Use the smallest practical:
 - (1) Field size.
 - (2) Exposure time.
 - (D) Consider the use of high kV technique to minimize skin dosage.
 - (E) Restrict fluoroscopy to indicated studies:
 - (1) Studies of dynamics and spatial relationships or guidance in spot filming details.
 - (2) Fluoroscopy is not a substitute for radiography.
 - (F) Protect patient gonads and fetus or embryo from useful beam to extent possible.
3. Personnel protection
- (A) Only needed persons should be in the room.
 - (B) All but the patient must wear protective aprons (at least $\frac{1}{4}$ mm Pb equivalent) unless measurement shows scatter intensity levels are 5 mR/hr or less.
 - (C) Operator's hands
 - (1) If they must be in the useful beam, the beam must be attenuated by the patient plus a lead rubber glove of at least $\frac{1}{4}$ mm lead equivalent.
 - (2) Such exposure must be only under exceptional circumstances.

* At least one state regulating code specifies 5R/m as an upper limit.

Medical Fluoroscopy

There are fluoroscopy radiation hazards to both personnel and patients. Personnel must be near patients during examinations, so they are exposed to scatter and leakage radiation. Substantial primary radiation dosage may be delivered to the operator's hands, which are often involved in positioning and other operations during fluoroscopy. The patient may be subjected to unnecessary irradiation as a result of excessive beam intensity from poor machines and excessive exposure. The use of image intensifiers helps yield greater information but normally does not greatly reduce patient dosage; the use of some automatic image intensity control units may inadvertently actually increase the beam intensity.

Fluoroscopy involves inherently greater patient dosage levels than radiography; therefore, its use is justified only by specific information yield (dynamics, spatial relationships, and spot film radiography of critical details) or by the absence of alternative radiographic facilities.

In this section we shall describe radiation safety recommendations more or less peculiar to fluoroscopy (Table 6) and briefly discuss some more hazardous machines and practices still occasionally encountered.

EQUIPMENT. As shown in Table 6(A) NCRP recommendations refer to both the x-ray beam itself and its primary and secondary shielding. Although the reasons for most recommendations are obvious and consistent with those of Table 2, some items merit particular comment.

Specific figures are given for the maximum allowed beam intensity. Both tube filtration and TSD are involved; if recommended values are employed the intensity will automatically be proper.

The primary beam shield must be adequate to protect the fluoroscopist. Actual measurements must be carried out with no patient in place to check the adequacy of the lead glass shield. During this measurement a large field must be used to envelop the entire cutie pie ionization chamber, with the chamber contacting the shield. The beam limitation requirements must be carefully checked, as any overlap of the shielding material by x-rays exposes the fluoroscopist's head to the dangerous useful beam.

Scattered radiation originates mainly in the patient. It may reach the doctor from both below and above the table, and appropriate shields are recommended (Fig. 2). Lead aprons, of course, absorb much of such scatter. However, since they do not cover the entire body, additional shields are desirable at the scatter source.

Cumulative timers are manually set switching units which shut off the fluoroscopic beam or energize a buzzer after a preset time of actual x-ray exposure, such as 3 or 5 such minutes. They must be

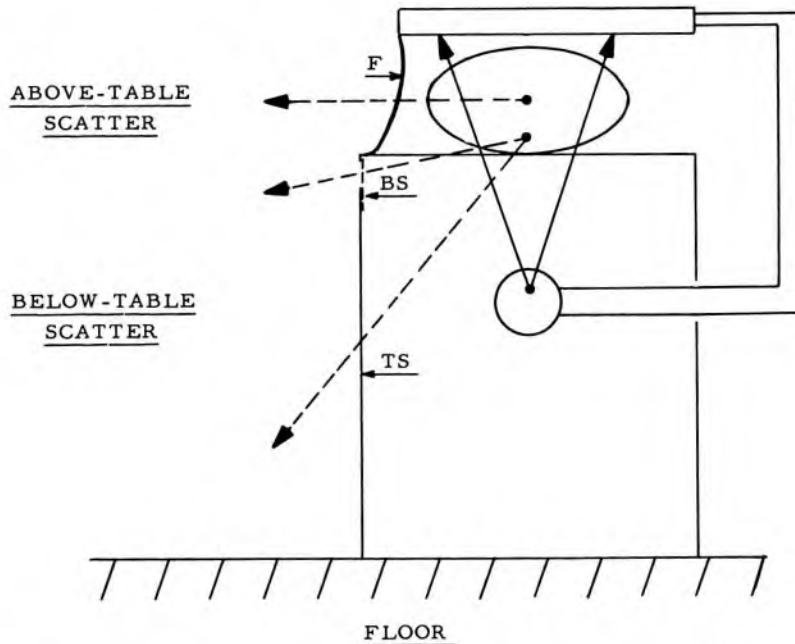


Fig. 2. Scattered radiation in fluoroscopy and equipment shielding arrangements. F—lead rubber slats hanging from fluorescent screen assembly. BS—bucky slot metal shield. TS—table assembly shield.

reset before the x-ray beam can be restored. Consequently, the operator is automatically reminded that his examination may be unduly prolonged. Although experienced fluoroscopists normally require no such reminder and usually recognize the desirability of such protection of the patient, beginners almost always benefit from such reminders.

Mobile fluoroscopy normally cannot be performed in a sufficiently dark room. Fluoroscoping under such circumstances jeopardizes dark adaptation, so ordinary fluoroscopy is inappropriate in mobile units. Good image intensifier units, preferably with television, are therefore recommended.

USAGE GUIDELINES. Even the best medical fluoroscopic equipment is hazardous to patients and operators unless used by or under the direct supervision of a physician properly trained in fluoroscopic procedures. Other more specific requirements relating to protection of both patients and personnel are given in Table 6.

Protection of the patient involves control of beam intensity, field size, and duration. An inexperienced or visually unaccommodated fluoroscopist is tempted to increase mA unduly to increase screen brightness. An upper beam intensity limit of 10 R/m is therefore set as an operation guideline. If adequate brightness is not achieved with reasonable beam intensity the fluorescent screen sensitivity should be checked against that of a new screen. Other items in Table 6 are reasonably obvious but are all essential to controlling patient exposure. Proper training of fluoroscopists is essential to reduce patient dosage.

Scattered radiation levels near patients receiving fluoroscopy can approach 500 mR/hr. There is, consequently, a serious potential whole body irradiation hazard to all persons in the room, and especially the operator. The use of lead aprons provides attenuation to substitute for permanent protection barriers which are impractical in fluoroscopy. The busy fluoroscopist may appropriately wear a somewhat heavier apron ($\frac{1}{2}$ mm instead of $\frac{1}{4}$ mm lead equivalent) than persons more remote from the patient.

Early fluoroscopists tested kV by inserting their hands in the x-ray beam and examining the image. Many sustained serious skin injury as a result. Modern competent fluoroscopists do not ordinarily receive exposure from the useful beam. When they must work with their hands in or near this beam, however, they do so on the *far* side of the patient and only with $\frac{1}{4}$ mm lead equivalent gloves for additional protection.

SOME MORE HAZARDOUS SITUATIONS. Some obsolete machines and practices involve increased radiation hazards. Machines are occasionally still seen in which the fluoroscopist's face may potentially be struck by the useful beam. One type is the hand-held fluoroscopic

viewing screen, which is freely movable. Other machines have screens which may be tilted or even fully retracted with the x-ray beam still turned on. All such machines are extremely hazardous to the operator.

Some x-ray examinations were carried out fluoroscopically in the past but are generally done radiographically nowadays. For example, simple fluoroscopy of the chest delivers substantially greater patient dosage (about 100 mrem/sec versus 40 mrem *total* for a radiograph) with less information yield than radiography. Fluoroscopy used to be preferred by some orthopedic surgeons because it yields immediate indication of the progress of surgical procedures. Rapid developing systems now provide radiographs promptly, however, with much greater detail visibility and reduced hazard to all concerned.

Other X-ray Applications

We have thus far discussed the most common medical applications of x-rays. We now consider two low kV applications as well as industrial radiography.

LOW KILOVOLTAGE X-RAY MACHINES. Reference has already been made to the high outputs of contact therapy x-ray machines having low filtration and very short TSD values (Chap. 7). Low filtrations are achieved with very thin glass or beryllium windows. The beryllium window tubes are particularly dangerous because beryllium transmits low energy radiation better than any other window material known. Outputs of the order of a million R/m may be produced near the target, when the tube is operated at 50 or 60 pkV. One can only stress the need for careful calibration and use of such potentially dangerous devices.

Low filtration x-ray tubes are also used for x-ray diffraction work. To obtain roughly monochromatic radiation such tubes have provision for target material selection and usually employ their own pumping systems. The most serious hazard is from irradiation of the skin and eyes during alignment and operation. With well-designed modern machines, shielding is relatively foolproof, although eye and skin injuries have been reported in the past.

INDUSTRIAL RADIOGRAPHY. Usually orthovoltage x-rays (150 to over 300 pkV) are employed to radiograph smaller, and supervoltage x-rays (1 to 35 MV) larger thicknesses of material. In all cases the installation must be well-shielded, and no personnel may remain in the room during exposures. There should preferably be no window, door, or ceiling openings in the room during machine operation.

Sometimes heavy work must be transported in and out quickly to expedite work, however, and permanent access openings are desir-

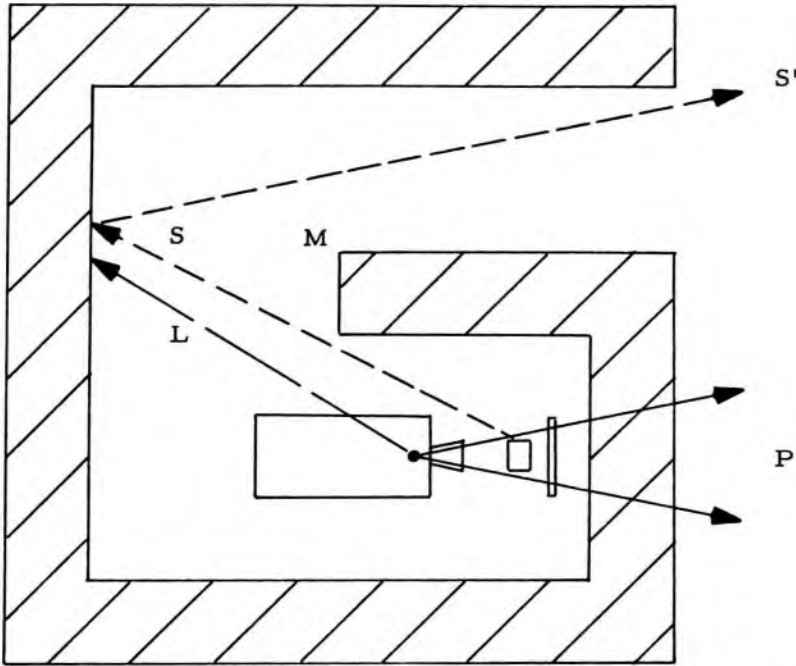


Fig. 3. Maze construction for radiation protection. For industrial radiography of large parts a heavy door is troublesome. The use of wall (M) introduces an indirect or "maze" access. Note scatter (S) cannot reach the room opening except after an additional scatter step, and with considerable distance attenuation. The intensity at the entrance is greatly reduced, so a heavy door can be omitted in many situations. The same is true for leakage (L). Mazes are widely used for medical supervoltage installations also, to reduce the required weight of treatment room doors.

able. The use of a maze (Fig. 3) is permissible if it is properly designed and later certified safe by a qualified expert.

In recent years large portable shielded enclosures have become available for industrial radiography and other x-irradiation. These enclosures are designed to provide primary and secondary shielding. Using these units, workers can function in the surrounding areas without excessive hazard while exposures are delivered within the enclosures. In addition to verification of the adequacy of such units for the particular use employed, suitable interlocks and other safety devices must be provided. If the use of such enclosures is contemplated, prior consultation with a qualified expert is recommended.

SEALED RADIOACTIVE SOURCES⁵

Radioactive materials are widely used for clinical as well as laboratory and industrial purposes. They are directly administered in clinical diagnostic studies and some therapeutic procedures, in so-called "unsealed" form. As previously discussed (Chap. 12), however,

therapeutic application of unsealed sources is limited by their uncontrollable spread to other than target tissues. Sealed sources, on the other hand, confine the source material while delivering the desired gamma or high energy beta radiation to the target area.

In radiotherapy, sealed sources are positioned both remote from and very close to tumors (*tele-* and *brachytherapy*). To produce useful intensity at suitable treatment distances teletherapy sources must be very active (for example, 1,000 to 10,000 curies of cobalt-60). Consequently, sources must be carefully shielded; the source assemblies also must not release significant amounts of highly radioactive powders.

Sealed brachytherapy sources are of much lower activity (milli-curie versus curie levels), so special housings are not required. However, they are necessarily moved about hospitals frequently and handled by many different people under often unfavorable conditions. Potential hazards therefore exist of substantial gamma irradiation to personnel and others. In addition, one may potentially lose or damage fragile sources, with the hazard of radioactive contamination of persons and areas.

This section discusses recommended procedures for both teletherapy machine and brachytherapy source usage. Other sealed source applications are also discussed. Later sections cover procedures relating to unsealed sources, special emergencies, and administrative record keeping.

Teletherapy Machines

Teletherapy units require beam control devices and procedures similar to those of high energy x-ray therapy machines they are designed to replace. In addition, problems of their own arise from their use of potentially hazardous high radioactivity sources.

Table 7 summarizes the main NCRP recommendations for these machines. Several merit specific comments, given below.

PREPARATION AND INSTALLATION. Required protective barriers are generally similar to those of supervoltage machines producing beams of similar quality. Thus, cobalt-60 beams are attenuated similarly to those of 3 MV x-rays, whereas cesium-137 beams are more similar to those of 1.5 MV x-rays. Governmental (AEC) regulations strictly control the legal delivery of teletherapy sources. In general, it is advisable to submit all required forms, plans, and calculations well in advance of construction to avoid delays and possible expensive room alterations.

“Beam interceptors” are special heavy shields provided on some rotational or “isocentric” machines to greatly reduce room concrete thickness and costs (Fig. 4). The shields are mechanically linked with

TABLE 7. RECOMMENDATIONS FOR TELETHERAPY INSTALLATIONS

A. Preparation and installation

1. Protection barriers are generally similar to those of 1 to 3 MV x-ray machines.
 - (A) Specific AEC licensure is required, however, of room plans, usage, and physician's qualification before installation of a source.
 - (B) Beam interceptor can reduce necessary primary barrier thicknesses; it should
 - (1) Intercept all scatter within $\pm 30^\circ$ from the central ray,
 - (2) Reduce the transmitted intensity so secondary barrier thickness is adequate, and
 - (3) Be provided with means to prevent deflection or removal of the barrier.
2. Interlocks, patient communication, and beam direction limitation are the same as for x-ray therapy machines.

B. Checkout is always by QE, with written report and copy to AEC.

1. Installation check and calibration requirements are the same as for x-ray therapy machines.
2. Equipment
 - (A) Protection housing
 - (1) Leakage radiation, beam OFF. At 1 meter, 10 mR/hr *max*, 2 mR/hr *average*.
 - (a) Any one reading is taken over up to 100 cm² area.
 - (b) Field survey readings: 14 points (see Figure 4). At initial installation and after source replacement.
 - (2) Leakage radiation, beam ON. Less than 0.1% of useful beam intensity, except for collimator region.
 - (a) Check not required if information is available on exact prototype.
 - (b) Measure with collimator closed completely as possible and collimator blocked with absorber equivalent to adjacent housing.
 - (c) This requirement is waived for installations used only for whole body irradiation.
 - (3) Fire protection
 - (a) The housing should be constructed to preserve shielding integrity in case of fire.
 - (b) The source capsule construction should minimize the probability of escape of radioactive material.
 - (B) Radioactive source
 - (1) Must resist breakage, with a double-welded seal.
 - (2) Leakage of radioactive material.
 - (a) Accessible surfaces of the housing must be wipe tested after source installation and at least every 6 months thereafter.
 - (b) In OFF position, wipe with moistened cotton or filter paper and assay latter for contamination.

(continued)

TABLE 7. RECOMMENDATIONS FOR TELETHERAPY INSTALLATIONS (continued)

- (c) If transferred activity exceeds $0.05 \mu\text{Ci}$, take action to limit contamination and notify authorities.
- (C) Beam defining apparatus must transmit not more than 5% of the central ray beam intensity. Supplementary loose blocks, etc. may transmit more.
- (D) Exposure shutter control
 - (1) Operation
 - (a) In ON position, source must be aligned with collimator.
 - (b) Shutter must work well in any housing orientation for which it is designed.
 - (c) Ordinary shutoff: must be designed so failure is highly unlikely.
 - (d) Shutoff must be automatic in power or similar failure.
 - (e) Emergency manual shutoff should involve minimum personnel exposure.
 - (2) Switches
 - (a) Automatic timer.
 - (b) Beam energization possible only from control panel.
 - (c) Interlock control on door.
 - (d) Warning devices must indicate beam ON and OFF, on both control and housing.

C. Guidelines for the user

1. Guidelines for x-ray therapy also apply here, relating to calibration, observation of patient and panel, patient holding, and a machine locking device to prevent unauthorized use.
2. Emergency procedure, in case shutter remains stuck in ON position, must be established and posted at the control panel. See NCRP Report Number 33 for sample instructions.

D. Source disposal

1. Source removal and disposal must *both* be performed only by persons specifically licensed to perform these operations.
2. Source disposal must be properly noted in the installation records, and notification given to appropriate authorities.

the housing so they always intercept the useful beam.* They must be large enough to intercept patient scatter within a cone of $\pm 30^\circ$ as well. Such radiation is both intense and relatively hard; scatter at greater deflection angle is much softer and adequately absorbed by secondary barriers. The interceptor must attenuate all but about 0.1 percent of the primary and hard scatter beam to permit the use of secondary barriers only.

* On occasion machines have been installed with the housing free to rotate so the beam no longer strikes the interceptor. This is hazardous unless the barriers struck are specifically designed for the unattenuated primary beam.

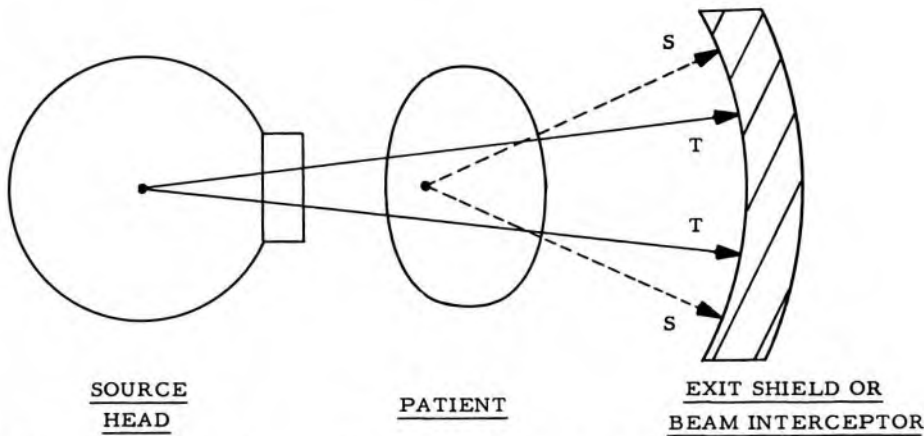


Fig. 4. Use of an exit beam interceptor in teletherapy. Note radiation both directly transmitted (T) and scattered through less than 30 degrees (S) is intercepted by the exit shield. The degree of attenuation must be adequate to permit use of secondary barriers only, in a manner analogous to what is done in fluoroscopy.

CHECKOUT. The AEC has strict rules concerning both initial and periodic checkout of teletherapy machines. Copies of all QE reports must always be kept on file and available for inspection.

The source housing must meet exacting requirements. To protect personnel, leakage radiation limits must be low even with the beam OFF. These limits are set adequately low for necessary work with patients, but it is evident the teletherapy room is no place to loiter or work regularly for prolonged periods after treatment hours.

The 100 cm² area figure in leakage measurements (Table 7) is selected to facilitate the use of a cutie pie instrument and because narrow higher leakage beams are not likely to consistently strike the same tissues of exposed personnel. The fourteen test points represent a departure from previous recommendations of N.B.S. H. 73⁶ involving an improved selection of points.⁷ Their spatial distribution is described in Figure 5. Leakage requirements with the beam ON are similar to those for high energy x-ray beams, generally. Some machines are used only for whole body irradiation; the question of housing leakage in the ON position is then less relevant, and the above requirement is waived.

Very special source containers are required to prevent contamination from the enormous activity sources employed in teletherapy. High pressures can develop inside sources from decomposition of small amounts of salts and water as well as from elevated temperatures. Capsules are hence required to be of special construction, with double-welded seals. Strict testing is required at all stages of production and assembly.

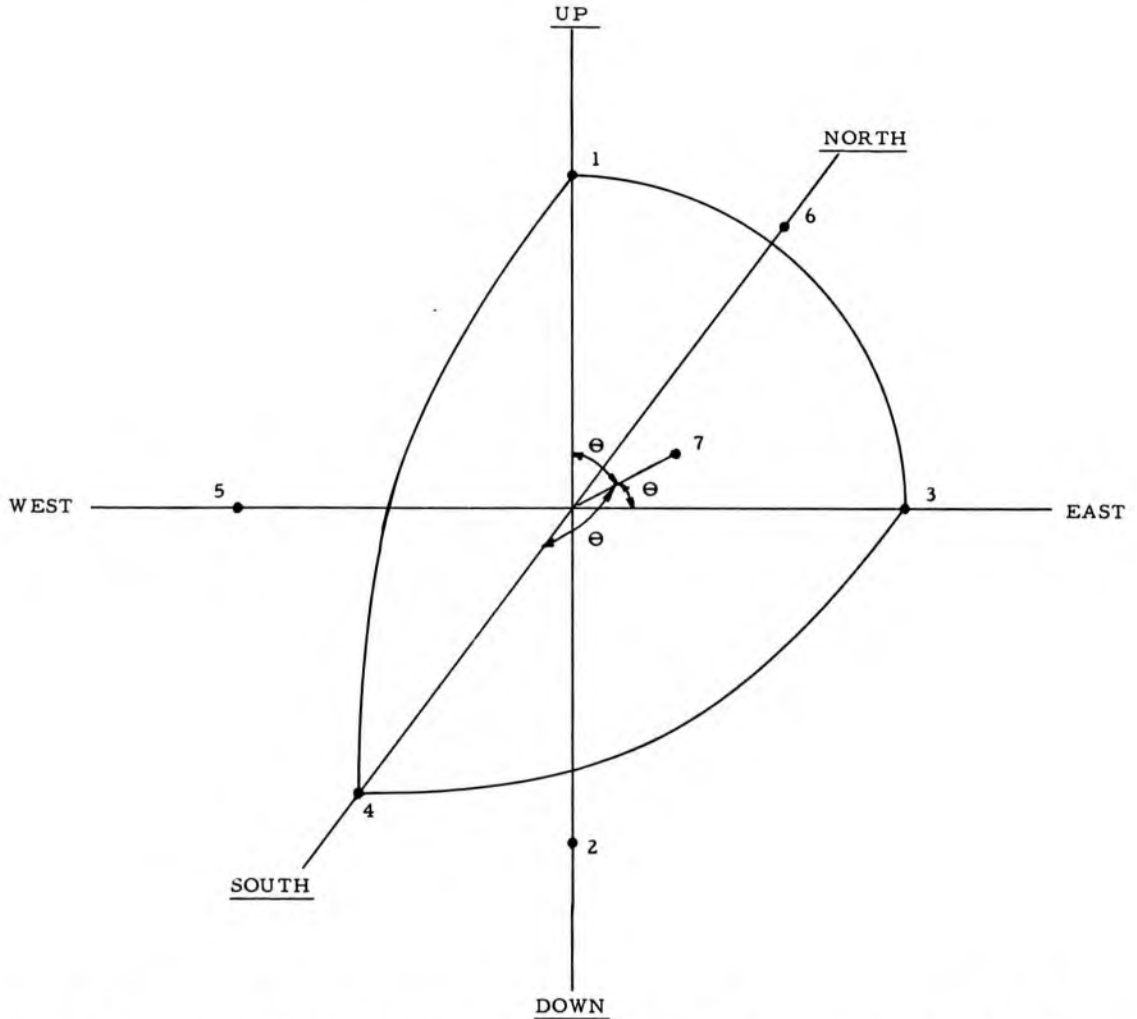


Fig. 5. Location of checkpoints, all 1 meter from source, for teletherapy housing leakage in off position. Six basic points: four on horizontal equator 90° apart (3 to 6) and two at corresponding poles of 1 meter radius sphere (1 and 2). Eight additional points: each is in the geometric center of the eight spherical triangles determined by the above six points (example: point 7). For field installations: these 14 points are sufficient for initial and subsequent tests. For factory tests: 12 other points are also required midway between the basic six points on the three great circles they determine.

The great radiation dosage received by source material accelerates chemical effects such as the corrosion of cobalt-60. As a result, some radioactive powder is inevitable in even metal teletherapy sources. If the seals fail some powder may escape as air- or gravity-borne dust. Since the cobalt-60 has enormous specific activity, so has the powder. For example, specific metal source activities of 75 Ci/g are common, so 1 mg of oxide powder could have 50 mCi of cobalt-60!

The necessity to detect contamination promptly means wipe tests of accessible surfaces are required before using a new source as well

as at least every six months. Measurement of more than 0.05 microcuries *total* removed activity is strong evidence the source is defective. "Accessible surfaces" generally include collimator mirrors, jaws, and plastic cover surfaces. Often the collimator must be removed to conduct the test properly.

The shutter mechanism must also meet high standards. In some machines the source is rotated to the ON and OFF position at the periphery of a wheel assembly (see Figure 1 of Chapter 16). Should the mechanism function improperly the final ON source position could be misaligned with the collimator, shifting the treatment field unreproducibly.

Since radioactivity cannot be turned off, the shutter must be as dependable as possible. Usually a positive spring mechanism is employed to return the source position or shutter to the OFF position even if the power or other mechanisms fail. Nevertheless, shutters *have* become "stuck" in the ON position on occasion, requiring emergency manual shutoff. This should be possible through a rear approach to avoid the need to enter the useful beam. Instructions should clearly explain the procedure as well as the quick removal of the patient from the room. As a prompt warning to the operator, lights or other devices are required to indicate any failure of the shutter to close.

USAGE GUIDELINES. The guidelines applicable to radiation therapy (Table 1) apply here as well. The procedure for emergency shutter closure is also very important, as is regular wipe testing for leakage. Although beam intensity tends to fall uniformly (about 1.1 percent per month for cobalt-60), calibration or spot checks should still be carried out regularly. The central ray intensity of a teletherapy unit varies greatly with field size, much more rapidly than that of a similar energy x-ray machine, because contributory scatter of the source is a significant part of the beam. Intensity data are hence required for several field sizes to avoid significant error. For example, one cobalt-60 unit varies 10 percent in beam intensity as the field size increases from 35 through 250 cm².

SOURCE DISPOSAL. Normally a cobalt-60 source is replaced after a maximum use of five years when its activity is one half the initial value. This is still a highly active source (250 to 2,500 Ci) with extreme potential handling hazards. Special lead "pigs" are used for transferring such "spent" and new sources; these are essentially duplicate housings with provisions for very rapid shielded transfer. As in all production, acquisition, use, and transfer of fission-produced radioactivity, special AEC licensure is required to perform transfer and disposal work.

TABLE 8. OPERATIONAL STAGES IN CLINICAL USE OF BRACHYTHERAPY SOURCES

1. Transfer of sources from storage and preparation for use on patients.
2. Transfer from preparation bench and application to patient.
3. Irradiation of the patient.
4. Removal of sources after completion of treatment and transfer back to preparation bench.
5. Removal of sources from applicators, cleaning, and transfer from preparation bench to storage space.

Brachytherapy Sources

Unlike those used in teletherapy, brachytherapy sources are handled by personnel and transported regularly around the hospital. These activities inevitably result in some irradiation of doctors, technologists, nurses, as well as bystanding patients and visitors. Table 8 lists five operational stages in the clinical use of brachytherapy sources.

After brief consideration of basic radiation hazards in brachytherapy, we shall discuss procedures to minimize these hazards at the various operational stages.

HAZARDS OF BRACHYTHERAPY. Ordinary brachytherapy hazards arise primarily from external irradiation. Almost all these sources emit energetic gamma rays, so exposed persons may receive whole body irradiation. Those actually handling sources can also receive considerable skin dosage, especially to the hands. Beta rays which are hazardous to the eyes are also sometimes present; strontium-90 eye applicators as well can deliver significant local skin or eye irradiation if used improperly.

Should source containers be damaged, some of their contents may irradiate local skin and eye areas. In addition, internal exposure may result from the ingestion and inhalation of radioactivity. Table 9 lists the most common brachytherapy materials, their form, and principal hazards.*

Soluble radioactive materials are particularly hazardous inside the body because they may be absorbed and distributed to various organs. All modern source materials are, fortunately, insoluble, but sources containing soluble radium and cesium-137 salts are still occasionally used. Radium chloride and bromide salts are particularly dangerous because radium is a bone-seeker. (The presence of soluble

* Data from NCRP Report No. 33.⁸

TABLE 9. CONTAMINATION HAZARDS OF RADIONUCLIDES USED IN BRACHYTHERAPY*

Radio-nuclide	Form	Typical Encapsulation	Active Material Solubility	Principle Contamination Hazards	Remarks
Radium-226	RaSO ₄	Double seal, Pt-Ir alloy	Insoluble	Ingestion and inhalation	Not absorbed from gut. Retained in lung and pulmonary nodes depending on particle size.
Radium-226	RaCl ₂ or RaBr ₂ [†]	Same as above	Soluble	Ingestion and inhalation	Absorbed readily into body. Some retained by bone.
Cobalt-60	Alloy	Stainless steel	Insoluble	Ingestion and inhalation only if fragmented.	
Cesium-137					
New	Ceramic microspheres	Double seal	Insoluble	Ingestion and inhalation	Not absorbed by gut. Pulmonary retention depends on particle size.
Old	Ce ₂ SO ₄	Double seal	Soluble	Ingestion and inhalation	Ingestion and inhalation hazards.
Iridium-192	Metal with stainless steel or Pt sheath	Stainless steel or platinum	Insoluble	Very low	
Gold-198	Metal	Platinum sheath	Insoluble	Very low	
Tantalum-182	Metal	Platinum sheath	Insoluble	Very low	
Strontium-90 (eye applicators)	SrCO ₃	Glazed silicate plus Al foil cover	Insoluble	Ingestion and inhalation	

* See text for further information.

[†] Obsolete materials, found in very old sources.

or fixed radium may be revealed by radon-222 in exhaled air. The presence of insoluble RaSO_4 is not so revealed because over 97 percent of the radon is trapped in the salt.) Some older cesium-137 sources contain soluble sulphate (Cs_2SO_4). Cesium-137 is far less radiotoxic, microcurie for microcurie, but *any* accidental contamination with significant quantities of soluble radionuclides must be treated by an expert.

Although less hazardous than the soluble salts, insoluble powders are still hazardous in the body. (These include radium sulphate, cesium-137 "microsphere" ceramic powders, and strontium-90 carbonate.) When *ingested* these are all eliminated in time, but catharsis is advisable to speed up their removal, and monitoring by external counting is recommended to follow the process. The fate of *inhaled* insoluble powders depends on their particle size. Particles larger than about 10 microns are quickly removed from the lung by ciliary action and then swallowed; however, smaller particles may penetrate the lung alveoli and remain there for days, months, or, in some cases even years.

All other materials in Table 9 are essentially chemically resistant metals, and their hazards hence arise only from proximity of the intact source.

Table 10 summarizes recommendations for the control of radiation hazards in brachytherapy. We now consider some of these recommendations.

STORAGE AND IDENTIFICATION OF SOURCES. A storage and handling area should have limited access to avoid loss of sources and unnecessary exposure to outsiders. However, adequate space is recommended to permit efficient operation, with ventilation (preferably directly to the outside) to remove any radon gas from radium sources. Shielding of stored radium sources must be adequate in all directions, usually accomplished by the use of a commercial lead "radium safe" with added 2-inch lead bricks as needed. In addition, returned radium is sometimes left soaking for many hours before return to the safe. Significant dosage may be delivered to persons on both sides and in areas above the radium if additional lead protection is not provided.

Identification of sources can be a problem because engraved serial numbers are very tiny and it is hazardous to read them at close range. It is advisable for sources of different strengths and total activity to be readily identifiable by length and shape, to facilitate recognition at a reasonable distance. Grouping similar sources in separate safe drawers also helps in this regard. An additional benefit arises from the fact that only a fraction of the total sources is unshielded while a drawer is open.

TABLE 10. RADIATION SAFETY PROCEDURES AND DEVICES
IN USE OF BRACHYTHERAPY SOURCES

A. Storage and Identification of Sources

1. Area

- (A) Ample space and ventilation; controlled access.
- (B) Normally, storage and handling areas adjacent to facilitate shielding during transfer to and from safe.

2. Shielding

- (A) Adequate in all directions.
- (B) Provision for shielding returned and soaking radium as well as ordinary storage.

3. Arrangements

- (A) Sources grouped to both facilitate identification and expose worker to only part of total supply during use of area.
- (B) Sources marked for quick identification at reasonable distance.
- (C) List on wall of complete source supply, with full description of each source.

4. Available means to check source strengths.

B. Handling Equipment

1. Basic essentials

- (A) L-block, forceps, clamping devices; needle and capsule threader.
- (B) Lead transfer containers.

2. Afterloading and cycling devices.

3. Some optional devices: magnetic, pneumatic, and servo devices.

C. Checkout of Sources (upon arrival and periodically thereafter)

1. Accounting with strict permanent record keeping of *all* source use and return.

2. Visual examination to detect bending or other damage.

3. Assay for strength and distribution of activity in source.

4. Contamination checks

- (A) All sources: wipe tests for free radioactive material.
- (B) Radium: radon leak tests.

D. Usage (see Table 9)

1. Transfer and preparation for patients

- (A) Proper personnel training and tools to minimize exposure.
- (B) Mark sources to facilitate identification.
- (C) Avoid heat sterilization of sources containing powder (cesium-137 and radium). Carefully check those which have ever been heat sterilized. Verify chemical action of sterilizing solutions. Consider ultrasound methods.

(continued)

TABLE 10. RADIATION SAFETY PROCEDURES AND DEVICES
IN USE OF BRACHYTHERAPY SOURCES (continued)

2. Application to patient
 - (A) Train all personnel in hazards involved.
 - (B) Where possible, provide shielding covered with sterile drapes in relevant directions.
 - (C) Where not possible, rely on speed and distance to minimize exposures.
 - (D) Forceps must be provided for handling sources, and their use strictly enforced.
 3. Precautions during treatment
 - (A) For higher activities mark bed and chart with appropriate information and warnings.
 - (B) Required patient segregation depends on mCi hours, location in patient, and type occupancy in adjoining beds (see Table 12).
 - (C) For higher activities wristband on patient with appropriate data.
 - (D) Patients with removable sources should not be permitted to leave clinic or hospital. With non-removable sources, procedure depends on nuclide and mCi level.
 4. Removal of sources from patient
 - (A) Observe same precautions as during insertion.
 - (B) **RETAIN ALL LINEN, DRESSINGS, CLOTHING, AND EQUIPMENT IN CUBICLE OR ROOM UNTIL ALL SOURCES ARE ACCOUNTED FOR.**
 - (C) In cleaning sources, be careful to prevent source damage or loss into plumbing system.
 5. Source accounting
 - (A) There must be an assigned custodian of sources responsible for maintaining permanent records of all source removal and return.
 - (B) Records must include
 - (1) Source order, with date, date received, patient, physician, hospital, or department.
 - (2) Source issued, type and identification of applicator, number of sources, total mCi, date and time of issue, and person to whom issued, with signature.
 - (3) Returns, with date expected, actual date of return, signature of person certifying complete return of sources.
 - (4) Periodic inventory of all sources.
- E. Disposal
1. Renewal or credit on new sources may involve reincapsulation of several damaged sources, etc., shipped to a supplier. He will generally advise on packaging for the transfer.
 2. Disposal without credit on new sources generally requires contacting a commercial disposal firm. Source must be packaged appropriately.



Fig. 6. Radium storage and handling facilities at Hahnemann Medical College and Hospital. A. General area is well lighted, with a ventilating hood on the right. B. Use of special brachytherapy tools. Shown is use of a special vise to hold a radium source vertical during string attachment. A special threader used to insert the string and long forceps help assure hands are several inches from the source at all times.

HANDLING EQUIPMENT. Certain devices are basic essentials for handling radioactive sources. These are illustrated in Figure 6, a photograph of the radium storage and handling area at Hahnemann Hospital. The L-block protects the operator from both horizontal and oblique gamma rays with 5 cm thick lead. This is about 4 HVL's for radium or cobalt-60 gamma rays, providing an attenuation of 16 times. Thus, the worker's torso receives roughly 5 rems for 75 rems received by the hands at the same distance from the source. Where desired, the torso dosage can be further reduced by the use of additional lead bricks.

Forceps are an absolute necessity to protect the handler's fingers. Many types are available, specifically designed to facilitate manipulation and with assured good grip of sources without excessive pressure. Vises, chucks, and other devices are needed for many applicator manipulations. They, too, should be designed to hold sources firmly without excessive pressure. Threaders are essential for threading needles and capsules, to reduce handling exposure not only to the

hands but the eyes as well. Lead transfer containers should have long handles to assure substantial distance of the radium from the torso of the carrier and should offer at least a half inch of lead shielding. Heavier carriers should be on wheels.

The greatest external irradiation hazards in brachytherapy arise from handling active sources during their application to the patient. This may involve time-consuming procedures, often repeated after radiographic check of source distributions in or around the lesion being treated. Even skilled operators may receive substantial local and whole body irradiation under these circumstances.

"Afterloading" and "cycling" are two techniques used in some larger institutions to reduce radiation exposure. Afterloading involves the use of hollow non-radioactive containers which are inserted at leisure in the patient. Once the placement is radiographed and found to be satisfactory, radioactive sources are added *afterward*, relatively quickly, and sometimes at bedside. The resulting reduction in dosage to the operator and his assistants may be quite substantial because of reduced exposure time. Cycling is a special afterloading technique involving only one source at a time. When the tube has been placed in the patient, the source is mechanically inserted by remote control. Large activities are used (0.5 to 3.0 curie) with short treatments (minutes versus days).

Other special grasping and servo devices have been used. They offer some advantages, but their use has not yet become very general in hospitals.

CHECKOUT OF SOURCES. Strict accounting of sources is essential, and responsibility must be centralized in a single person. A complete written record must be maintained of all release and return of sources. Any persons charged with the return must be trained in the procedures involved. Losses must be reported immediately to designated persons responsible for emergency procedures, and steps taken to prevent the spread of any radioactive contamination. Sources, particularly those used interstitially, may become bent, flattened, or otherwise damaged. They should be routinely examined and removed from use if defective. When first received, sources should be checked as to their strength. In addition, radium and some other radioactive materials may sometimes be improperly distributed along the active length of the source. Autoradiographic checks quickly reveal this.

Source leakage checks are essential for materials of more than 30 days half-life to avoid potential contamination of patients. The measurement technique used must be capable of detecting 0.005 microcuries of the radionuclide involved. The preferred method varies with the type of source. Most sources are tested by either washing in

detergent and counting the solution or by a wipe method followed by counting. Radium sources are tested by measuring radon leakage. Leaky sources must be removed from service and either disposed of or re-encapsulated. Tests are required immediately after delivery, if source damage is suspected, when contamination is measured, and at intervals not exceeding six months.

USAGE GUIDELINES. Control of radiation exposure during applicator preparation is facilitated by the provision of shielding and handling devices. Operators, however, must be well-trained to assure proper use of the equipment. Heat sterilization is damaging to seals of some sources, and some chemical sterilizers may also be destructive.

As previously indicated, the application of sources to the patient can be potentially hazardous to personnel. In the past, operating room pressures have sometimes led to direct manual handling of sources to expedite procedures. The use of forceps, speed gained from skill, and, where possible, shielding can greatly reduce personnel exposure. When practical, afterloading techniques are likely the most effective way to reduce exposures at this stage.

During treatment the sources are at best in or on a patient and at worst may become dislodged. The patient, his chart, and the bed must all be appropriately marked to announce that radioactive materials are present and to provide necessary instructions to physicians and nurses attending the patient. The question is frequently raised of required isolation of the patient under treatment. A guideline is that adjacent patients should not receive more whole body dosage during the treatment of up to 5 days than the occupational guideline figure of 0.1 rem. Table 11 indicates required distances for five common

TABLE 11. RELATIONSHIP BETWEEN mCi HOURS AND DISTANCE REQUIRED TO ASSURE LESS THAN 0.1R EXPOSURE TO NEARBY PERSONS FOR FIVE COMMON RADIONUCLIDES*

Milli- curie Hours	Required Distance from Unshielded Sources for 0.1R or less Exposure				
	Radium (Feet)	Cobalt-60 (Feet)	Cesium-137 (Feet)	Iridium-192 (Feet)	Gold-198 (Feet)
10	0.9	1.2	0.6	0.7	0.5
30	1.6	2.1	1.0	1.3	0.9
100	3.0	3.8	1.9	2.3	1.6
300	5.1	6.5	3.2	4.0	2.7
1,000	9.4	11.9	5.8	7.4	5.0
3,000	16.3	20.5	10.1	12.7	9.0
10,000	30.1	37.6	13.5	23.2	15.8

*Data from N.B.S. Handbook 73, Table 7.

radionuclides and assumed mCi-hr treatments. For many treatments it is evident that the usual roughly 6 feet separation of patients is considerably less than required for 0.1 R total exposure, and exposures closer to 1 R are common during cervix cancer therapy. The neighbors in both the same and adjacent rooms should therefore preferably not be young persons and women of childbearing age. (Hospital partitions are usually effectively penetrated by most gamma rays used in brachytherapy.) As a general rule patients with removable sources should be kept at the clinic or hospital to prevent irradiation of other persons if sources leave the patient. Careful monitoring of sources by hospital personnel is generally essential.

Source removal upon termination of treatment is usually performed in the patient's room. Special care to account for all sources is required under these relatively unfavorable circumstances. It is generally a good idea to have a second person radiation monitor the patient, all linen, dressings, clothing, and equipment while the removed sources are being returned to the preparation bench. The author has observed several instances in which sources were left behind while their empty applicators were returned. Sink drains, wastebaskets, and the floor should also be checked. Finally, sources are rather small and can easily be washed down drains unless care is taken. Once a source moves beyond a trap, retrieval may involve an expensive plumbing job.

Disposal of sources involves many legal as well as technical considerations, and consultation with a qualified expert is often desirable.

Other Sealed Sources

We shall now consider some other applications of gamma- and beta-emitting sealed radionuclides.

GAMMA SOURCES. These include sources used for calibration and teaching, industrial radiography, and special irradiators.

Very low activities are often used for calibrating sensitive meters, and levels of below 1 mR/hr at 5 cm distance are usually considered safe for portable sources. Readily identifiable larger sources may be mounted inside an instrument, provided accessible surfaces have levels below 10 mR/hr and the sources are clearly identified.

In general, if there is much chance of a worker's receiving 100 mR in one hour inside an area, access must be carefully controlled. Sources producing greater intensity than 1 R/hm (1 R/hr at 1 meter) are very dangerous and should not be handled without both remote handling apparatus and special shielding.

Industrial radiography is sometimes performed using high activity sources (commonly ^{60}Co). Room shielding and access considerations are similar to those of industrial x-ray radiography. In addition, the source must be housed in a shielded container, with a positive action mechanism to automatically return the source between exposures. In this regard the situation is quite similar to teletherapy. The use of any unshielded high activity ^{60}Co source is very hazardous and not recommended.

Special irradiation setups have been built, usually employing multiple high activity ^{60}Co sources. For example, multiple cylindrical ^{60}Co sources have been used to irradiate large animals in an open field as well as patients in a special appropriately shielded treatment room. (This is essentially a crossfire technique.) Generally, exposure is initiated by raising the sources into position above ground from buried shielded housings. Fairly uniform whole-body radiation can be administered with good design of source locations.

Smaller units have also been built, also using peripherally placed rod sources, for small animal and specimen irradiation. In these units the sources are usually mounted in heavily shielded containers into which the preparation is inserted. Remote control devices are recommended in these devices for inserting and removing preparations because of the very high gamma intensities at the irradiator opening.

BETA EMITTERS. Thickness gauges employ a beta-emitting strip located below a moving sheet of plastic, rubber, paper, or similar material. A detector is placed above the rolling sheet stock, so the indicated count rate varies in response to average material thickness. This can be a very sensitive and convenient system, but the unit must be designed to minimize any beta-ray hazard to workers. Contamination from the source (such as a ^{90}Sr - ^{90}Y foil) must be prevented.

High energy beta sources like those of ^{90}Sr may produce significant penetrating bremsstrahlung radiation. For example, ^{90}Sr eye applicators can yield several mR/hr at a meter and should be stored with some awareness of their potential hazard.

UNSEALED RADIOACTIVE SOURCES

Radiation hazards naturally increase with the activity handled. One speaks qualitatively of three general situations: low level or "cold," "warm" or "hot," and "very hot." Medical tracer studies are essentially low level, involving microcurie amounts except for somewhat larger stock solutions; little difficulty ordinarily arises in such

work. Therapeutic uses are warm or hot, being associated with larger activity and hence exposure levels. Far greater activities are present in very hot applications, which include plants for processing nuclear reactor fuel rods, radiopharmaceuticals, and other such products; in some such activities special robot handling and transfer systems are required for safety. We shall consider only the low level and warm or hot types, and particularly medical applications; the same principles are applicable to other uses involving similar activities and radionuclides.⁹

Unsealed sources are normally employed in solution or suspension form. Hazards therefore include area contamination as well as internal and external irradiation of personnel. In addition, the remote possibility exists of administering the wrong material or activity to patients, so some sort of assay of administered material is desirable. Finally, when higher activities and/or more hazardous radionuclides are used, one must be prepared for unlikely but possible accidental or emergency exposures of personnel.

Our coverage therefore includes the following topics: storage and handling facilities, assay, general laboratory rules, special problems in radiotherapy, disposal, and emergency procedures and decontamination.

Storage and Handling Procedures

When present in millicurie amounts, all gamma-emitting radionuclides may require special shielding. In clinical usage the requirements are less with unsealed than with sealed sources because, typically, solutions are generally used up promptly or decay quickly. Also, many have lower Γ and HVL values than those of radium, cobalt-60, and cesium-137. Special shielding against beta rays is usually unnecessary, with an exception arising in the storage of energetic radionuclides like phosphorus-32 and yttrium-90, etc. These can produce penetrating bremsstrahlung, so storage of multicurie amounts in low Z containers is usually desirable.

Perhaps of prime importance is a suitable selection of the storage location (Table 12). It should be close to the handling area, to minimize the chance of spills and gamma exposure, and have convenient access to a sink with a good drain. (A good drain is one with as rapid flow as possible to the main sewer without likelihood of backup into other areas.) Furthermore, access should be limited to minimize exposure of other personnel and the general public. Remoteness from areas occupied regularly by other people is always helpful if it can be arranged.

TABLE 12. TYPICAL MEDICAL HOT LABORATORY REQUIREMENTS

A. Storage and Handling Area

1. Storage

(A) Isolated area preferred, free from traffic.

(B) Adequate lead shielding for amounts and types stored.

2. Handling

(A) Near storage area and sink, with good drain to sewer.

(B) Nonabsorbent surfaces, including work and sink surfaces, and preferably walls and floor as well, to facilitate decontamination.

(C) Normally special arrangements for ventilation unnecessary even with I^{131} . For handling gaseous and more volatile materials, fume hood or other system may be required.

B. Facilities and Accessories

1. Handling devices: tongs, forceps, remote pipettor, shielded syringes, rubber apron.

2. For spills: metal trays and special backed absorbent paper.

3. Glassware: racks and pipette washer with drain.

4. Emergency items: special garments, respirator, shower (for handling dangerous materials like radium salt, etc.).

5. Special hot storage: for excreta, contaminated articles.

C. Instruments

1. Survey: G-M and cutie pie instruments are basic.

2. Assay: ion chamber instruments are desirable, also reference standards for counting.

D. Administration to Patients

Procedures should preferably be performed in hot lab area.

Handling procedures require more detailed discussion. They involve those associated with preparation operations and with persons or animals receiving radioactive materials.

BENCH HANDLING. Both external and internal exposure hazards may exist. When millicurie quantities of gamma emitters are involved, remote pipetting and transfer devices are useful during initial dilution and subsequent solution transfer. Lead shielded storage enclosures are useful when practicable. When small quantities of very active solution are handled, syringes and solution containers may be shielded also.

Even moderate activities present potential contamination hazards. The use of nonporous wall, floor, and table surfaces helps greatly in

TABLE 13. GENERAL RADIOACTIVITY LABORATORY RULES

A. Hot Lab

1. Minimize external exposure by fast work, use of remote handling devices, and shielding.
2. Minimize contamination.
 - (A) NO SMOKING, EATING, OR DRINKING IN AREA.
 - (B) Use gloves in handling high specific activity liquids; wash promptly with soap and water.
 - (C) Use metal trays and absorbent paper.
3. Area monitoring of benches, floors, and other likely locations, also fingers, hands, etc. Use thin-window G-M tube for beta-ray detection.
4. Personnel monitoring with film badges generally and hand and finger badges for some work. Pocket dosimeters for more hazardous procedures involving gamma emitters.

B. Other Locations

1. Administering of material: spilling is a potential hazard. It is preferable when possible to do this in the hot lab, to minimize transfer distances.
2. Radioactive samples: not normally a serious problem. NEVER PIPETTE BY MOUTH.
3. Measurements: avoid contamination of wells and other sensitive detectors.

decontamination procedures. (Wooden or other porous materials should be avoided or covered appropriately.) During all transfer operations one should use stainless steel trays and special absorbent paper with a water barrier backing. Care should be taken to avoid spillage or ingestion and the touching of areas known to be contaminated. Wearing rubber gloves prevents the deposit of radioactive material on and in crevices of the skin; gloves can be either readily washed after use to remove any surface contamination or disposed of. Laboratory clothing should also be provided to minimize the transfer of radioactive material on street clothing. When very high activities are employed, plastic overalls, helmets, and other special garments are used, but this precaution is not normally required in medical applications. Specific suggestions regarding general laboratory rules and procedures are given in Table 13.

PATIENT AND ANIMAL HANDLING. Some gamma- and/or beta-irradiation hazard is present during the administration of radioactive solutions. Containers of solutions should be handled as little as possible by personnel, and always with rubber gloves when using greater activities.

In both human and animal work, tracer studies generally involve no significant hazards after administration. This is no longer true, however, in radiotherapy (see below). Animals given large doses pose hazards from feces, urine, and other contamination routes. Cages should be in restricted areas with good drainage, and attendants should be provided with detailed instructions and sometimes radiation monitoring. Periodic area monitoring may also be required.

Assay

One should always check that received shipments are of the proper material and activity. Although most radionuclide suppliers are quite reliable, errors have occasionally occurred in the past. A shipping error can be particularly regrettable in radiotherapy, but spoiled tracer studies also waste time and irradiate patients needlessly.

Millicurie quantities should at the least be checked grossly for gamma versus beta activity (i.e., iodine-131 versus phosphorus-32). The activity of gamma emitters can be checked to fair accuracy by a simple cutie pie measurement at a known fixed distance. (Measurements are usually taken to the geometric center of the ionization chamber.)

A preferable method employs a more sophisticated instrument and source geometry, the British National Physical Laboratories unit, which is commercially available. In fact, such units are indispensable for checking some of the newer short-lived pure gamma emitters, like technetium-99m, which are made in the laboratory using appropriate cows. In addition, an aliquot sample should be tested in a reliable well counter with a suitable window setting. A good multi-channel analyzer is best of all for verifying the radionuclide involved.

Pure beta- and even some gamma-emitting radionuclides may be accurately assayed by beta counting of aliquots. End window, gas flow, and 4π counters have all been used with reference standard sources. (These are usually longer-lived sources whose beta-ray energies approximate those of the radionuclides being assayed.) Backscatter and absorption errors must be controlled by proper sample drying and mounting; the details of such procedures are beyond the scope of this book.¹⁰

General Radioactive Laboratory Rules

Table 13 summarizes basic rules for hot laboratory areas. Though consistent with common sense and the content of our previous discussions, certain rules merit specific comment.

The ingestion of radioactivity is an ever-present hazard around a hot laboratory; the rules against smoking, eating, and drinking

should be strictly enforced. Mechanical pipetting is accurate and quite easily learned, so mouth pipetting of radioactive solutions is inexcusable.

Area monitoring should be performed frequently and spill areas cleaned up promptly to avoid smearing material onto the skin and clothing. The use of metal trays and wax paper or plastic backed absorbent paper in likely spill areas greatly facilitates cleanup. Inexpensive thin window beta-ray G-M detectors are sufficiently sensitive to detect most radionuclides. It is useful and reassuring for the handler to check accessible skin areas and clothing with a G-M unit, particularly following more hazardous procedures. Personnel monitoring is generally required.

The unnecessary transport of radioactive solutions should be avoided. It is best for this reason to administer material to patients in the hot lab, but when this is impossible, transfer should be carefully executed to avoid spills or container breakage on the way. The use of a cart with appropriately shielded bottom-heavy containers is recommended.

The contamination of radiation detectors can be an expensive problem. The use of a commercially available special metal insert shield is recommended, especially when solutions of longer half-lives are measured (such as ^{60}Co , ^{137}Cs , ^{51}Cr , etc.).

Special Problems in Radionuclide Therapy

Relatively large amounts of radioactivity may be administered in radionuclide therapy; from 5 to more than 100 mCi are common. In addition to gamma-ray hazards, those from contamination are also present. We may conveniently consider these hazards at three stages: preparation and administration, patient handling, and emergencies such as special surgery or death.

PREPARATION AND ADMINISTRATION. Following assay the material must be prepared for administration. A suitable syringe or gravity feed device may be used for intravenous or intracavitary insertion. Gamma-ray exposure to personnel is minimized by expediting delivery and shielding the container holding the material. Testing for tight tubing and needle connections and the use of absorbent paper help minimize contamination. Skill and teamwork are, of course, essential.

When the insertion is complete and the wound, if any, closed, all possibly contaminated articles, including containers, tubing, syringes and needles, dressings, linen, absorbent paper, etc., must be collected and removed. The entire area must then be checked for contamination and cleaned up. Contaminated articles may be cleaned

and then either stored until radioactivity is negligible, disposed of directly, or removed by an appropriate disposal service.

PATIENT HANDLING. If a gamma emitter is administered the patient may serve as a source of external irradiation just as in brachytherapy, and similar labeling and isolation procedures are recommended (Tables 10 and 11.).

Contamination may arise from leakage from the wound and other body openings. Dressings, linen, and clothing must be saved for check and decontamination. Urine and feces are saved in some treatments for dosimetric purposes. Rubber gloves should be provided for handling such potentially radioactive materials and instructions provided. Written instructions are recommended, in the form of a Nurses' Manual, regarding excreta handling as well as the dressing of wounds and ordinary nursing procedures. Ordinary dietary, housekeeping, and maintenance activity can be safely carried out expeditiously. Before discharge, the patient should be thoroughly checked for surface radioactive contamination and all possible contaminated effects removed and processed. Usually simple instructions to the family are sufficient when the patient is discharged.

SPECIAL SURGERY OR PATIENT DEATH.*¹¹ If the patient requires emergency surgery, the surgeon must be notified of the amount, type, and site of the radioactive material. Preferably, the extent of any radiation hazard involved should be specifically discussed so any necessary precautions may be taken.

Should the patient expire with radioactivity still present, proper procedures must be followed to protect the pathologist and mortician.*

Special precautions should be taken during autopsy:

1. In general the radiation protection supervisor should be present or at least consulted before starting.
2. Loose radioactive fluid should be removed by suction, followed by sponges held with forceps.
3. Heavy or multiple layers of surgical gloves are advised for beta shielding, with goggles to protect the eyes.
4. Rotation of pathologists is advisable, if practicable.
5. Dissection should be performed with an 8 inch forceps and long-handled scissors. Removed organs should be stored a few feet from people.
6. The area and all tools and garments should be decontaminated upon completion.

* Recommendations are taken from N.B.S. H. 65, which is currently being revised; some may be changed in the new handbook. The main points, however, are relevant and not likely to be changed basically.

The body may be released for embalming as long as it is certified to contain less than 30 mCi of radioactive material. If more is present, required precautions must be specified by the radiation protection supervisor. Cremation is permissible within specified annual radioactivity totals for the crematorium (200 mCi of ^{131}I , 200 mCi of all other radionuclides).

Complete records must always be provided, showing clearly the type and amount of radioactivity present and how it was placed there, autopsy arrangements, final accounting of radioactivity disposal (including the amount sent to the mortician), and whether burial or cremation was performed.

Disposal of Radioactivity

In ordinary work, disposal involves aqueous solutions and animal excreta primarily. Radioactive surgical specimens and cadavers, however, may present special problems.

Radioactive solutions and excreta may ordinarily be safely flushed down drainage systems. (The term "batch" disposal is often used in this connection.) The use of direct drains is important to reduce the chance for contamination should local drain blockages occur. The average effluent concentration must not exceed MPC levels for those persons in the environs. Hospital water usage is generally very great, so considerable dilution occurs. The storage of materials until they decay to low levels is usually possible and strongly recommended when larger activities of moderate half-life materials are to be disposed of.

The disposition of human bodies with radioactivity has been discussed above.

The U.S. AEC permits *no* incineration of radioactive materials without specific AEC license to do so.¹² The death of a radioactive animal can therefore present a problem. Commercial disposal by AEC- or state-licensed firms is possible at a cost depending primarily on the bulk volume of material. This disposal method becomes expensive when large animals are involved. Burial is also permissible if performed in an AEC- or state-approved manner. This usually involves digging rather deep holes and is normally not a very practical solution. Even licensed incineration requires maintaining careful records, and incinerated amounts are generally limited to low microcurie amounts each week.

EMERGENCY RADIOACTIVITY PROCEDURES

The unexpected can happen in even the best-run department, with radioactive as with any other hazardous materials. The greatest

worry concerns loss of material with the possible contamination and exposure of people. The twin objectives are to confine contamination to the local area and to protect personnel.¹³ An establishment, to best handle such situations, should have immediately available a qualified expert for advice; a radiation protection supervisor (RPS) on the staff, with executive authority to act and the basic background to do so intelligently; a medical team knowledgeable in radiation injury and its treatment; and security, maintenance, and housekeeping personnel clearly instructed by top management to follow instructions of the RPS without delay.

In medical work, serious emergencies may arise primarily from teletherapy or brachytherapy sources damaged in fires, floods, explosions, or other unusual circumstances. In some non-medical situations, spills may involve high activities and long half-lived materials; such an occurrence is far less likely in medical laboratories. The following discussion is intended to cover such very unusual circumstances.

Immediate Action

When trouble is suspected, one must locate the material as quickly as possible. Retracing transfer steps is useful in searching for lost radium. If radioactive powder is lost, it is essential to find every suspect person and check for activity on body, shoes, clothing, bedclothing, food trays, toolboxes, automobiles, etc.

One must close off any contaminated area, a necessary procedure not only to protect outsiders but also to prevent the spread of radioactivity. This means blocking off the area and preventing airborne contamination as well. Fans and air conditioners should be turned off and doors and windows closed. All persons not needed in the area should be promptly evacuated after rapid but careful survey shows them to be free of contamination; all possibly contaminated persons, however, should be detained.

The radiation protection supervisor, qualified expert, medical team, and administration should be notified immediately. Generally, local or state authorities must also be notified if any potential public health hazard exists.

Cleanup of Contaminated Areas

If significant amounts of contamination exist, cleanup should never be attempted without expert help. The procedure should be planned carefully, as in the handling of virulent organisms. Contaminated areas must be clearly marked off until cleaned up.

A vacuum cleaner with a micropore filter should be used first to pick up most of the material. (The individual carrying out this hazardous procedure should wear suitable respiratory protection.) The use of sticky tapes is another useful method. Only after this operation should scrubbing and mopping begin, carried out with constant monitoring. Coveralls, shoe covers, and multiple layers of surgical gloves should also be worn. Persistent significant contamination may require removal of patches or floor and other surfaces.

Film badges or other personnel monitors should be worn by all persons involved in the decontamination procedure.

Possibly Contaminated Persons

The following basic rules apply in dealing with persons who may have significant surface or internal contamination.

1. Retain the individual and carefully survey his skin, hair, shoes, and clothing.
2. Carefully remove all contaminated clothing and place in a disposable container.
3. Bathe contaminated areas gently with a mild soap. Avoid increasing skin permeability. Special suspect areas are nostrils, ears, body folds, scalp, and under nails. Use cotton swabs in cavities.
4. After contamination is removed, give the person clean clothing.*
5. An immediate medical check is suggested. It should include a complete medical occupation history, chest roentgenogram, CBC, hematocrit, and routine urinalysis. Blood samples, urine, and the feces should be collected for a full 72 hours. The radionuclide should be identified. If appropriate, do whole body counting, radon breath samples, etc.
6. Obtain reports of all film badge or other monitors worn by the exposed person during the exposure.

ADMINISTRATIVE ASPECTS

In most states, ionizing radiation sources are subject to special regulations which may require source registration as well as the maintenance of other records. The AEC also requires special licensure for procurement and use of most radionuclides. These are enforced by the AEC except in those states which by mutual agreement have taken over the licensing function. Users must follow regu-

* Suggested levels of "significant contamination" are specified in Table 6 of N.B.S. Handbook 92.¹⁴

lations with scrupulous care to avoid criticism which can jeopardize important programs and even reflect adversely on the reputation of the operator.

We shall therefore consider essential aspects of the two most common sources of administrative difficulty, AEC licensure and radiation records.

U.S. AEC Licensure

As a result of its strict enforcement of basic safety rules, the U.S. AEC has achieved an exemplary record in its own manufacturing and research operations. This policy has been extended to its control of the general use of radionuclides. Such control results from the power to withhold reactor-produced materials. Non-governmental persons must be specially licensed to manufacture, use, transport, or dispose of AEC-controlled radioactive materials.¹⁵ Recently, the AEC has transferred administrative authority to some states, but regulations remain substantially the same. We shall consider licensure for the use of radionuclides primarily, since it is of the most general interest.

OBTAINING A LICENSE.¹⁶ A license is granted to a particular person to possess stated maximum amounts of particular radionuclides in particular chemical form for particular uses in a particular location for a specified time (usually two years). If the licensee changes his position, the license becomes void; neither he nor the management is authorized to obtain more materials, and those already received must be properly accounted for to avoid citation. To regain licensure, the licensee must reapply at his new location, and the management in turn must obtain a suitably qualified new person and apply all over again.

Two pitfalls, the improper transfer and careless disposal of licensed material, should be avoided. Once the licensee receives material, he is held accountable by the AEC or state inspector for every microcurie. Transfer is permitted only to another properly licensed individual, and accurate and self-consistent records are required at both installations. Similarly, all disposal records are carefully scrutinized by inspectors. Incineration of radioactive specimens, samples, and carcasses is of particular concern. No radioactivity may be knowingly incinerated without specific license to do so. Since most medical teaching institutions routinely incinerate dead animals, inspectors may be quite specific in their queries as to disposal of radioactive experimental animals.

The AEC has several application forms of which the medical types are considered here. AEC-482 is used for physician registration,

and AEC-313 MC, AEC-313, and AEC-313a are used for licensure application. The AEC Licensing Guide explains and illustrates all these forms and is very helpful in the application for licensure.¹⁷

Broad permission may be granted any physician with a state drug license to carry out certain routine diagnostic studies. Form AEC-482 is first executed and submitted by the doctor. When approved and registered, the returned copy is essentially a general license. Studies are limited to those for which small activities of pre-packaged individual doses may be used: ¹³¹I (thyroid uptake), ¹²⁵I and ¹³¹I (blood and plasma volume determinations), ⁶⁰Co and ⁵⁸Co (intestinal absorption of cyanocobalamin), and ⁵¹Cr (red blood cell volume and survival). These are now well-established studies involving negligible occupational or patient hazards even without special handling facilities and training.

Form AEC-313 is basic to all other licensure. An alternative form AEC-313 MC is useful to expedite application for well-established diagnostic and therapeutic procedures. Both forms request the following information concerning:

1. *The user* and his radiation training and experience.
2. *Arrangements*: radiation detection instruments and their standardization; facilities and equipment; film badges, dosimeters, etc.; radiation protection program; and waste disposal.
3. *Institutional approval*.
4. *Radionuclides, their form, and the purpose of contemplated use*.

Formerly only AEC-313 was employed. It is still used for all but well-established medical applications. Form AEC-313 MC was introduced a few years ago to reduce the number of license amendments required in expanding nuclear medicine programs. As long as he has adequate experience in *one* of several uses in a given category, the AEC-313 MC licensee is free to perform *all* other studies in the category. (Uses in a given category are similar in radiation hazards to operator and patients.)

Form AEC-313a is a supplemental form and must be completed in addition to AEC-313 or AEC-313 MC if human use is contemplated. The proposed diagnostic or therapeutic application must be stated with detailed data regarding patient dosage schedules, assay method, and other relevant information. Details are required concerning the physician's training, with written certification by the qualified preceptor. This usually involves experience in work related to the proposed application. If new or experimental work is involved, patient organ dosage calculations are also required as well as some comment regarding the potential benefit of the planned investigation.

INSTITUTIONAL LICENSURE. The AEC or state under certain circumstances is willing to delegate some of its enforcement and administrative responsibilities to responsible medical or other institutions. Under these circumstances an institutional committee carries the full burden of enforcing AEC regulations and in effect does its own licensing. Its proceedings and centralized records are available for periodic inspection, and the institution may still be cited for non-compliance.

The advantage of such an arrangement is local control, expediting the approval of new users, uses, and installations. It is especially helpful in extensive human pharmacologic evaluations. Unless the work volume is considerable, however, the administrative cost burden can become excessive, generally limiting such centralized licensure to large educational institutions.

KEEPING A LICENSE. As with a driving license, an AEC license is a privilege and can be withdrawn for major cause. This, however, is rarely done.

A visiting inspector is concerned primarily with general radioactivity housekeeping and records, including:

1. Personnel monitoring records, primarily film badges.
2. Area monitoring records, particularly floors, bench areas, doorknob and apparatus handles, wastebaskets, desks, etc.
3. Material records, not only shipment data but actual *microcurie amounts dispensed and disposed of, and how and when.*
4. General radiation levels, with area spot checks for gamma intensity and contamination.
5. Condition of monitor instruments and records of annual calibration.
6. Obvious non-compliance items, such as poor storage, handling, transfer, etc.

In addition, there are certain formal requirements for the proper use of radioactivity warning signs, posting of form AEC-3, and radioactive labels on bottles, etc.

Radiation Records

Besides the AEC, some state regulatory authorities require extensive radiation safety records covering personnel irradiation as well as radioactive and x-ray source records.

PERSONNEL. Personnel film badge records are a virtual medicolegal necessity for all users of ionizing radiation. Records are required for radiation workers and for other persons who may be reasonably expected to receive one fourth of the occupational MPD limit. It is

generally wise to provide a film badge for a reasonable period of representative exposure; if it shows levels to be very low, further film badges may be demonstrated to be unnecessary.

Records must show the employee's name, age, social security number, previous radiation history, and other relevant information. They must also include monthly or semimonthly exposure and quarterly and annual totals. In some more hazardous work, exposure to hands or other local body areas should be shown along with pocket ionization chamber or other personnel monitoring information. Upon his request, a worker must be provided with the total recorded dosage for his employment period. This information is often required by subsequent employers.

It is prudent to obtain, for future legal use if needed, a reasonably complete pre-employment health history, physical examination, and baseline CBC of a radiation worker. Of course, any occupational emergency exposure should be recorded along with all dosage records and medical measures taken. Radiation incidents must be reported in most states, and to the AEC if licensed radioactive sources are involved. Such reports may be immediate or deferred up to a week, depending on the seriousness of the hazard, in terms of exposure to individuals, radioactive contamination of air and water, shutdown time of the installation, and financial losses involved. (Further information is given in U.S. AEC Rules and Regulations, Part 20, Sections 20.403 and 20.405.)

RADIOACTIVITY. Record books must be kept of all received radioactive material, with dates, microcurie amounts dispensed, to whom or what, and for what purpose. The ultimate disposal of every microcurie received should be accounted for by decay and disposal. The method of disposal must also be stated, whether batch, burial, incineration, or removal by licensed carrier. Sealed sources must be regularly wipe tested and records made at least every six months. The AEC specifies 0.05 microcurie maximum removable accessible contamination for teletherapy, 0.005 microcurie for brachytherapy sources. If these amounts are exceeded, the source involved must be decontaminated or replaced by specially trained persons.

Finally, regular spot checks of laboratory surface radioactivity must be performed and data recorded for inspection. These measurements should be standardized so that a microcurie estimate can be made. It is surprising how high beta-ray levels can be reached in seemingly unlikely locations.

SOURCE REGISTRATION. In many states periodic registration (generally every 2 years) is required of all sources of ionizing radiation. X-ray machines as well as radioactive materials are included. Simple

as this requirement is, very often institutions discover they have acquired x-ray and other sources which were not reported to the radiation protection supervisor. Usually forms are provided for such registration, and relatively little effort is required to comply with this useful requirement.

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Calculation of Radiation Protection Barriers

In this chapter we discuss the basic principles of radiation protection barrier design.^{1, 2} Usually, a qualified expert is consulted in planning an installation. The future user, however, should be involved from the beginning in an intelligent collaboration with the QE and architect, assuring proper attention to other vital requirements, such as plumbing, electric and ventilation arrangements, and the over-all installation cost. A general background in radiation protection barrier design principles is essential for such collaboration.

Our coverage is necessarily limited to x-ray and gamma-ray beams.* Basic concepts and definitions are first discussed, followed by illustrative examples of four actual installations.

BASIC CONCEPTS AND DEFINITIONS

Figure 1 shows the basic situation in which a radiation protection barrier is used. Persons in an occupied area A must be protected from penetrating ionizing radiation originating at B, so an absorbing shield barrier C is interposed. By proper design the barrier reduces weekly dosage at A to values within the maximum permissible limits.

Thus, protection barrier design applies broad beam transmission data of various absorbing materials to given situations. Many technical aspects are also involved, relating to types of construction, materials, and radiation sources.

Computation of a Protective Barrier

Two steps are involved in the design of a structural barrier such as C in Figure 1. The first is to determine the required reduction in

* The design of barriers for nuclear reactors and high energy accelerators is beyond the scope of this book; our coverage of these devices will be limited to a brief discussion in Chapter 16.^{3, 4} As previously shown, alpha and beta rays are readily stopped completely and require no special structural barriers.

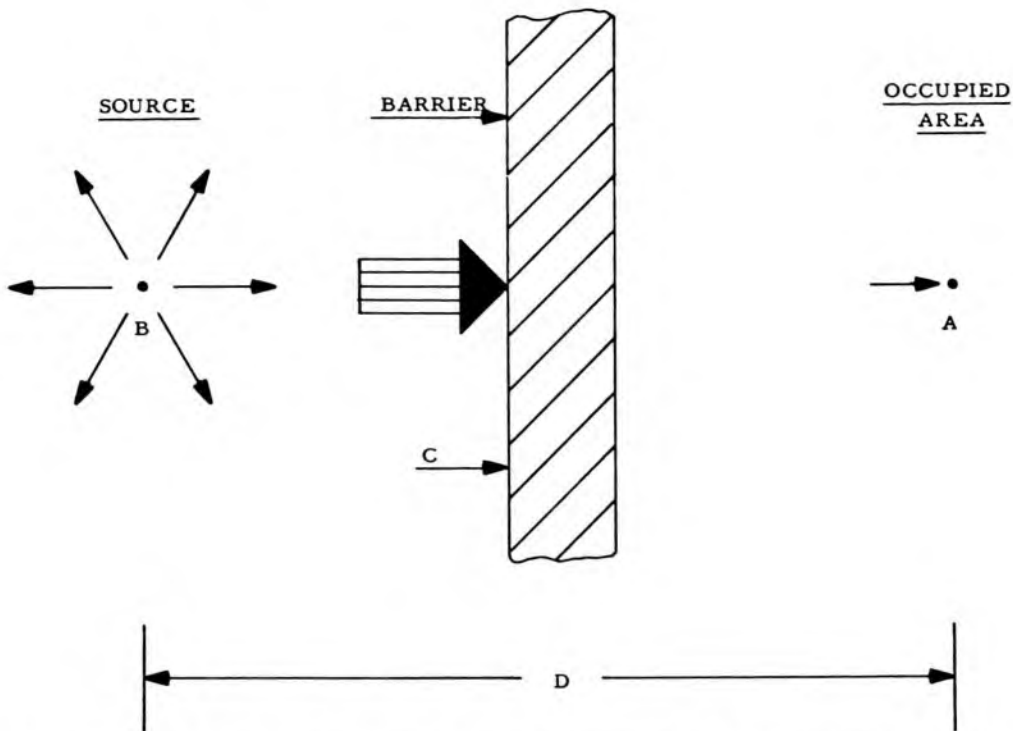


Fig. 1. Function of a protective barrier. Persons in occupied area A must be protected from penetrating radiation originating at B, so an absorber C is placed in between. If C is adequate, the weekly dosage levels at A are reduced to within permissible limits. The weekly dosage received by someone at A evidently depends on: the source intensity and its period of energization; the distance D; how often the source (such as an x-ray machine) is aimed at A; and how consistently the person remains at A.

exposure of the occupied area and the second, to provide the barrier thickness needed to provide this reduction. Practical design must also deal with many technical aspects of construction which affect the cost and shielding effectiveness of the final installation.⁵

REQUIRED BARRIER ATTENUATION (Table 1). As might be expected, the required attenuation depends on two quantities: the maximum permissible weekly dose for the persons involved (D_p) and the weekly dose delivered in the absence of a barrier (D_o). The permitted barrier transmission may reach a maximum value of T_m , where:

$$T_m = D_p/D_o \quad (15-1)$$

D_p is determined by the *type* of area occupancy. Recall the design figures for occupational whole body exposure is 0.1 rem/week. (This is taken as equivalent to 0.1 rad and 0.1 roentgen for photon beams in radiation protection work.) Levels only a tenth as great are generally permitted for exposure to non-radiation workers ("environs"),

TABLE 1. FACTORS DETERMINING THE REQUIRED ATTENUATION OF A RADIATION PROTECTION BARRIER

- A. Permissible weekly dosage at given location, D_p
 - 1. Type of occupancy, whether radiation workers or others.
- B. Dosage at this location without shield, D_o
 - 1. Dose at 1 meter delivered per week (or equivalent with x-ray units).
 - 2. Distance in meters.
 - 3. Use factor U .
 - 4. Occupancy factor T .
- C. The ratio of these, D_p/D_o , is used to determine the required barrier attenuation.

so the environs barrier may permissibly transmit only a tenth as much as a corresponding "controlled area" barrier.

The unshielded weekly dose depends on four quantities, listed in (B) of Table 1. The first two imply the dose a person would receive without a barrier present and with the beam aimed at him constantly while he remained in the area.

The second two are the use factor U (see Table 2) and occupancy factor T (see Table 3). The use factor is the fraction of the time a machine is turned on that a particular barrier is struck by the beam. For example, primary wall barriers usually have use factors of $\frac{1}{2}$ or $\frac{1}{4}$ but sometimes one. On the other hand, leakage and scattered radiation travel outward in all directions. Secondary barriers consequently have a use factor of unity regardless of the primary beam's direction. The occupancy factor refers to the fraction of the work week the given area is occupied. It could be unity for a location with a patient in bed or a sedentary desk worker; it is much less for an occasionally occupied rest room, closet, or narrow corridor.

For design purposes, it is the product UT which indicates the probability someone will be struck by the useful beam when it is turned on, and this product is actually used in design tables.

REQUIRED BARRIER THICKNESS. The needed thickness depends on the required attenuation and also on the broad beam attenuation characteristics of the barrier. As one would expect, the latter involves both the barrier material and beam spectrum. The situation is further complicated by multiple scattering within the barrier, so most broad beam transmission tables and curves are prepared from direct measurement rather than theoretical calculations. Considerable data are

TABLE 2. USE FACTORS FOR PRIMARY BARRIERS,
FOR USE IN PLANNING WHEN
COMPLETE DATA ARE UNAVAILABLE*

Barrier	Therapy	U, for Application: Radiography	Dental
Floor	1	1	1/16
Walls	1/4	1/4	1/4
Ceiling	1/16	1/16	1/16

*From N.B.S. Handbook 76, 1961, Table 3, p. 23.

TABLE 3. OCCUPANCY FACTORS, FOR USE AS A GUIDE IN
PLANNING SHIELDING WHERE ADEQUATE OCCUPANCY
DATA ARE NOT AVAILABLE*

Full Occupancy (T = 1)

Control space, offices, corridors, and waiting space large enough to hold desks, darkrooms, workrooms, shops, wards, nurses' stations, rest and lounge rooms routinely used by occupationally exposed personnel, living quarters, children's play areas, occupied space in adjoining buildings.

Partial Occupancy (T = 1/4)

Corridors too narrow for desks, utility rooms, rest and lounge rooms not used routinely by occupationally exposed personnel, wards and patients' rooms, elevators using operators, unattended parking lots.

Occasional Occupancy (T = 1/16)

Closets too small for future occupancy, toilets not used routinely by occupationally exposed personnel, stairways, automatic elevators, outside areas used only for pedestrians or vehicular traffic.

*Data from N.B.S. Handbook 73, 1961, p. 44.

available for standard materials, such as sheet lead and dense concrete of 147 lbs/ft³, but less for others.^{6, 7, 8, 9}

The design techniques involved will be illustrated by examples in the next section.

The selection of barrier thickness is usually carried out very conservatively, to over- rather than under-protect. Aside from humane reasons, there is the purely practical consideration of costliness and generally unsatisfactory results of adding protection to a finished room. Initial provision of ample barrier thickness may save much trouble later. At least four assumptions are made to assure conservative design:

1. The patient does not attenuate the useful beam.
2. An ample use factor applies, to allow for later expansion of activities.
3. The largest possible field and workload apply.
4. Both the type and extent of occupancy may change; an effort is made to anticipate future developments.

The experience of a QE is usually helpful in making these and other judgments involved in protection barrier design.

Technical Aspects of Construction

Certain principles help control costs and assure effectiveness of shielding construction. Some of the most important are listed in Table 4.

COST CONTROL. Three factors, the source-to-shield distance, occupancy, and ground location of the installation can greatly influence construction cost.

As a general rule, the closer the shield is to the source the less shielding material is required. As a rough approximation, the material volume is proportional to the *square* of the distance. Thus, 100 times as much lead is needed at 3 meters as at 30 cm. Storage of brachytherapy and unsealed radioactivity sources is usually carried out in compact enclosures. This permits the use of lead shielding at acceptable cost. For example, less than 400 lbs of lead is required for a certain radium storage unit because distances are only 10 to 20 cm. Lead, however, is impractical for a supervoltage installation with similarly penetrating radiation, where 3.2 meters distance is involved; a single wall might require 10 tons of lead to provide the same attenuation. The cost would be about \$6,000 for material alone in this wall! Fortunately, much cheaper materials are available for this purpose.

When possible, surroundings should be "controlled areas." Otherwise an additional tenth value layer (TVL) is required to reduce weekly dosage the required ten times to the "environs" limit. A tenth value layer is here defined as the thickness of absorber needed to attenuate the transmitted broad beam by a factor of 10. TVL values correspond to the hardened broad beam after much previous attenuation. Values for cesium-137, cobalt-60, and sealed radium are respectively 6.5, 8, and 9 inches of dense concrete and 2.2, 4.1, and 5.5 cm of lead. The TVL for 2 MV x-rays is very slightly less than that of cobalt-60; that of 6 MV x-rays is about 50 percent more, and of betatron beams even greater yet.

Regardless of the type of occupancy, reducing factor T can save on barrier material. Peripheral building locations are therefore

TABLE 4. PRACTICAL ASPECTS OF PROTECTION BARRIER DESIGN AND CONSTRUCTION

A. Cost Aspects

1. Distance to barrier: volume increases roughly as *square* of distance.
2. Occupancy of neighboring areas
 - (A) Type: a tenth value layer extra is needed for environs vs. controlled occupancy (6 to over 10 inches extra dense concrete for supervoltage rays).
 - (B) Degree: cf Table 3. Use of peripheral room location may be helpful.
3. Ground floor location is desirable.
 - (A) Critical in supervoltage and teletherapy
 1. Floor is always a primary barrier. Concrete 3 to 4 ft thick often needed!
 2. Enormous weights involved. Solid structural support required.
 - (B) Problem not serious for diagnostic x-rays but often serious for orthovoltage x-rays installations.

B. Special Shielding Problems

1. Assembly of lead sheets
 - (A) Wall joints: overlap is required of $\frac{1}{2}$ inch or twice the lead thickness, whichever is greater.
 - (B) Protect against sag or cold flow and mechanical damage.
2. Doors
 - (A) Avoid exposure to primary beams.
 - (B) Watch for leaks around door jambs, sills, and doorknobs.
3. Ventilation
 - (A) Louvers may be needed even in supervoltage installations.
 - (B) In teletherapy and supervoltage installations, mazes or baffles are generally preferred to effectively block direct scatter through openings.
4. Plumbing and electrical conduits are best located in the floor or in baffled or maze areas.
5. Services of a qualified expert are generally desirable for all but the simplest installations.

generally preferable for penetrating beams, especially supervoltage. Fences and warning signs and lights may be used to limit access to surrounding areas when it is impractical to provide full protection barriers. However, the safest procedure is to design for unlimited occupancy and to confine limited occupancy arrangements to exceptional situations where no reasonable alternative exists.

Ground floor locations are generally useful in orthovoltage and are *essential* in supervoltage installations. Specific floor shielding barriers may usually be omitted in ground floor locations. The saving involves up to $\frac{5}{8}$ " lead floor covering in an orthovoltage, 3 to 6 feet of

concrete in a supervoltage room, which are otherwise required to protect the room below.

Considerable difficulties may arise when any supervoltage installation is located above the ground floor. The 3 to 6 foot floor barrier is not only costly in material but also requires the sacrifice of ceiling space in the room below. In addition, a tremendous weight of the room ceiling, walls, and floor must be supported above ground, so an unusually elaborate structural base may be required. For example, a particular 6 MV linear accelerator room requires more than 300 tons of dense concrete shielding. Special footings are needed to support this much weight even at ground level. Far more elaborate and costly support structures are required for a room aboveground.

Other recommendations are given in N.B.S. H.73 and H.76.^{10, 11}

SHIELDING BARRIER CONSTRUCTION. Table 4(B) indicates several requirements peculiar to radiation shields. All lead joints and seams must be adequately overlapped, with the lead sheets touching if possible, to avoid passage of radiation through openings. Lead is a soft metal, so great care must be exercised to avoid damage in handling as well as later cold flow and tearing due to its softness and great weight.

Doors are especially difficult to shield. Even ingenious designs may be only partially effective, and leakage is generally measured around the door frame and doorknob. Doors are therefore poor primary barriers. Another pitfall is ventilating ducts or window openings. These should be located near the ceiling and away from any primary beam. In addition, lead louvers may be needed; they must provide the same attenuation as the secondary barriers they replace. In supervoltage rooms, mazes usually provide the most satisfactory solution; ducts can enter the room through the maze, near the ceiling; plumbing and electrical conduits are preferably placed in the floor. In other installations where ducts and pipes must penetrate lead barriers, suitable lead flanges and sleeves may be used to minimize leakage.

Lead and Dense Concrete (Table 5)

As indicated above, lead and dense concrete are most commonly used in barriers. We shall now briefly discuss these as well as some other materials of practical interest (see Figure 2).

LEAD. Sheet lead is provided in plates and in roll form, as well as bonded to cinder blocks or sheet plywood. Thicknesses range from 1/32 to more than 1/2 inch. It is used for rooms containing x-ray machines operated up to about 250 kVcp or 300 pkV. In such usage attenuation factors of 0.0001 are readily obtained with lead thick-

TABLE 5. PRACTICAL ASPECTS OF BARRIER MATERIALS

A. Common Materials

1. Lead: sheeting in plates or rolls and bonded to cinder block or plywood.
2. Dense Concrete: 2.35 g/cm^3 or 147 lbs/ft^3 .
 - (A) Usually cast in plywood forms with steel reinforcing.
 - (B) Also, *solid* dense concrete blocks—*not ordinary concrete blocks, which are hollow.*

B. Lead [Fig. 2(Top)]

1. Used as barrier material for attenuating:
 - (A) X-rays up to 250 kVcp or 300 pkV.
 - (B) Radium and unsealed radioactive materials being stored.
2. Basic advantage
 - (A) Pound for pound, best practical absorber* with least weight and bulk.
 - (B) This is because of high Z and density.
3. Disadvantages
 - (A) Mounting is expensive, both for bonded material itself and labor of assembly.
 - (B) Too costly for teletherapy and supervoltage.

C. Dense Concrete [Fig. 2(Bottom)]

1. Uses
 - (A) Supervoltage x-ray and teletherapy installations.
 - (B) Some lower kV x-ray installations, where space and barrier weight are not critical.
2. Usage
 - (A) Normally cast in forms, with steel reinforcement.
 - (B) Solid blocks: cracks between blocks may present problems. Use at least 2 rows, preferably with blocks staggered both vertically and horizontally.

D. Other Materials

1. Ordinary structural materials
 - (A) Siliceous and limestone materials: if solid, these approximate concrete, which they may replace if density corrections are properly made. They include brick, granite, limestone, marble, sand plaster, sandstone, and tile, but beware of hollow spaces and organic inclusions.
 - (B) Wood, Masonite, and other low density materials of organic origin are generally of little use for radiation shielding.

* Except for tungsten alloys, uranium, gold, and similar very expensive materials.

(continued)

TABLE 5. PRACTICAL ASPECTS OF BARRIER MATERIALS
(continued)

2. Window materials: lead glasses
 - (A) Ordinary lead glass
 1. Thin, used for diagnosis (1-2 mm Pb equivalent total).
 2. Laminated, for orthovoltage therapy and other applications requiring higher lead equivalent.
 - (B) Supervoltage: special dense cast glasses.
 - (C) All windows require special shielded frames.
3. Special materials
 - (A) Barytes plaster: diagnostic use, to avoid lead on walls and ceilings.
 - (B) Teletherapy and supervoltage: to reduce barrier thickness.
 1. Dense concrete: loaded with lead or barium salts to increase effective Z and density. Both cast and block form are used.
 2. Steel plates: used as primary barriers.

nesses less than $\frac{5}{8}$ inch. Another use of lead is for the shielding of brachytherapy and unsealed radioactive sources.

It is interesting to compare lead with concrete and other materials. Except for very expensive materials, such as uranium, gold, platinum, and certain tungsten alloys, lead generally provides a given attenuation with least bulk and weight, saving room space and making installation structural support easier. The efficiency in attenuation derives from the high Z of lead, which results in substantial photoelectric attenuation at photon energies as great as 250 to 500 keV, far above the corresponding 50 to 70 keV values for concrete. As a result, the lead curves (A) in Figure 2 rise much more steeply with operating kilovoltage than those of concrete (B). In addition, there is a region below 250 kVcp where lead thickness requirements are relatively low because of substantial photoelectric attenuation; corresponding concrete thicknesses become substantial at relatively low kilovoltages. To illustrate, consider three examples:

125 pkV x-rays (radiography): 0.0001 attenuation requires 10 inches of concrete, only 2.6 mm of lead.

300 pkV x-rays (orthovoltage): 0.0001 attenuation requires 15 inches of concrete, only 12 mm of lead.

Radium storage: 4 inches of lead protects as well as 23 inches of concrete.

CONCRETE. Lead is a relatively expensive material, and construction costs involved in its proper support are also great. This limits its use to applications where concrete or other cheaper substitutes are impractical.

Concrete is used for supervoltage rooms as well as for some lower kilovoltage x-ray installations in which space and weight are

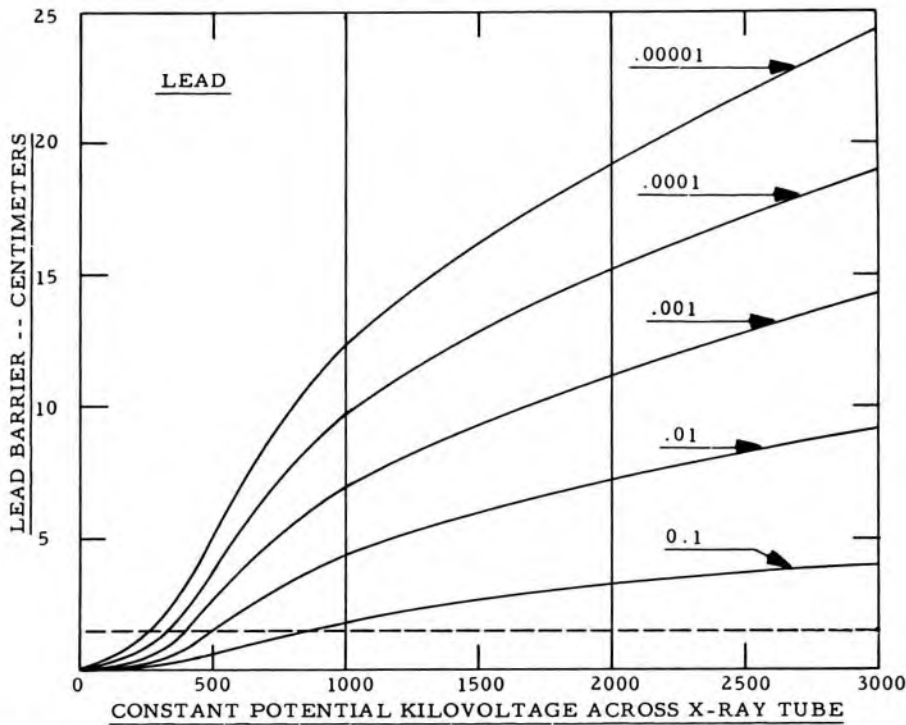
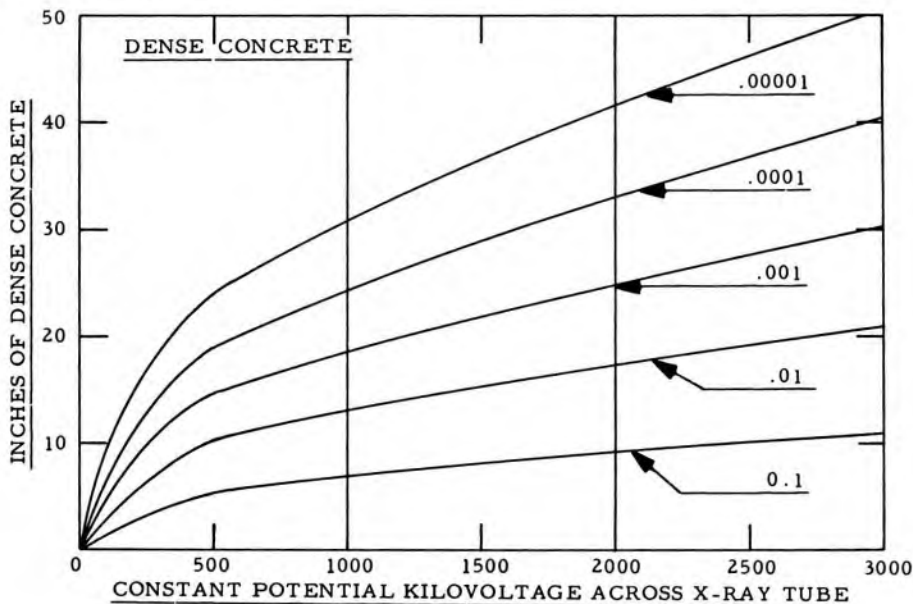


Fig. 2. Required barrier thickness vs. x-ray tube operating kilovoltage to attenuate broad beam to indicated transmission levels (.00001, .0001, etc.). Top. Lead barrier. Note substantial attenuation is achieved at 250 kVcp with only 1.5 cm thickness (dashed line). Lead is consequently quite economical for orthovoltage and diagnostic x-ray installations. Bottom. Concrete barrier. Note curves rise very rapidly and level off to become relatively flat above 500 kVcp. Very thick barriers are required even at 100 kVcp—more than half as thick as for orthovoltage beams! Concrete is most useful for supervoltage therapy installations because it costs much less than lead. It is normally not used at lower kilovoltages because the weight requirements are excessive.



not critical. Due to the low effective Z (between 13 and 14) of concrete, attenuation is predominantly by Compton scatter for x-rays generated above 150 pkV. This results in the need for substantial barrier thicknesses at even diagnostic kilovoltages, which increase relatively gradually with increasing kilovoltage [Fig. 2(B)].

Concrete is usually cast, carefully to avoid air holes, in plywood molds and reinforced with steel rods. Blocks are also used. Ordinary concrete blocks are hollow and less satisfactory for shielding purposes than special solid blocks. In any block construction, the cracks between blocks can in practice be only partially filled with mortar. For this reason the best barrier requires *two* layers, with blocks staggered both horizontally and vertically to avoid overlap of openings. Many masons are unaccustomed to these precautions, and close supervision is desirable. If blocks are placed without cement, cracks can be made narrow; ordinary masonry walls may be used to hold the loose blocks in place.

Other Barrier Materials [Table 5(D)]

We now consider several other absorber materials used in barrier design.

STANDARD BUILDING MATERIALS.¹² Ordinary building structures may provide useful shielding on occasion. Siliceous and limestone materials in existing walls can supplement added barriers and in some situations even make them unnecessary. In general, these materials attenuate reasonably similarly to concrete of the same weight per square foot. (Typical density figures are given in N.B.S. H.76, Table 15.) Existing brick, marble, and stone walls can be very useful in reducing new barrier costs, as can solid concrete floors. Plaster ceilings and walls are also sometimes helpful. However, it is always advisable to carefully verify the composition, thickness, and density of existing walls, ceilings, and floors when trying to save on new barrier costs. The writer has on occasion been grateful for a natural skepticism, when advisers made errors of up to 10 inches of concrete equivalent in their estimate of existing structures.

VIEWING WINDOWS. For diagnostic viewing windows one or two relatively thin sheets of lead glass provide the lead equivalent usually needed. In orthovoltage work a greater total thickness is required; several sheets may then be laminated to the required lead equivalent. In supervoltage and teletherapy installations solid high density lead glass blocks are sometimes used. These are quite thick, even when used as secondary barriers. They are sufficiently expensive that closed circuit television systems are popular alternatives.

All lead glass windows are provided with special frames of at least the same attenuation to prevent leakage around the edges.

SPECIAL APPLICATIONS. Special materials have been used as substitutes for lead and ordinary dense concrete for particular requirements.

In diagnostic applications a dense plaster can sometimes provide adequate wall and ceiling shielding, eliminating the need for lead. "Barytes plaster," which contains barium sulphate, is one such material. The high Z and density of this mixture enable it to attenuate diagnostic x-rays much more effectively than ordinary plaster.

In some supervoltage and teletherapy installations ordinary dense concrete barriers may be prohibitively thick, reducing the available room space excessively. Very dense concrete may be used to solve this problem. Barium sulphate and lead salts increase the concrete density and correspondingly reduce the required barrier thickness. In general, special concrete costs much more for both labor and materials, whether poured or assembled in blocks.

Sometimes steel plates are used to replace concrete in megavoltage installations. The steel density is about 7.8 g/cm^3 versus 2.35 g/cm^3 for dense concrete, and its Z is somewhat greater also. The result is that for 6 MV linear accelerator x-rays 1 inch of steel is equivalent to more than 3 inches of concrete. In one installation, for example, the requirement of a 5-foot concrete wall was met by use of a 2-foot-thick concrete wall plus a 1-foot-thick assembly of steel plates.

Specific Applications

The approach of Table 1 is basic to barrier calculations. In all applications the permissible weekly dose D_p is determined by the type of occupancy. Variations arise, however, among various radiation sources in evaluating D_0 , the weekly dose without shielding. In addition, the form of the published data also varies with the type of source.

Before considering the examples of the next section, we shall briefly describe the data used for three major categories of installations: x-ray machines, teletherapy units, and radioactivity storage.

X-RAY MACHINES (Tables 6 and 7). The combination of beam kilovoltage, wave form, and workload determine the appropriate x-ray shielding table. (Normal variations in filtration are ignored because they do not substantially affect the intensity of the most energetic photons in the primary beam. Only these photons penetrate the barriers significantly; the rest are effectively absorbed.) Constant potential and pulsating are the only two wave forms usually considered; more specific characterization is unnecessary for this pur-

TABLE 6. SHIELDING REQUIREMENTS FOR BUSY RADIOGRAPHIC INSTALLATIONS*

[W=1,000 ma-min/week at 100 kvp, 400 ma-min/week at 125 kvp, or 200 ma-min/week at 150 kvp.]

Distance from Tube to Occupied Area	Type of Barrier											
	For Controlled Areas						For Environs					
	Primary			Sec- ondary	Primary			Secondary				
UT	1	1/4	1/16	1	1	1/4	1/16	1/64	1/256	1	1/4	1/16
5 ft (1.52 m)												
Lead												
mm	1.9	1.4	1.0	0.5	2.7	2.2	1.7	1.2	0.8	1.2	0.8	.5
in.	1/16	1/16	1/32	—	3/32	3/32	1/16	1/16	1/32	1/16	1/32	—
psf†	4.4	3.3	2.3	1.2	6.3	5.1	4.0	2.8	1.9	2.8	1.9	1.2
Concrete												
in.	6.1	4.6	3.2	1.7	8.1	6.6	5.3	4.1	2.8	3.8	2.8	1.7
psf†	75	56	39	21	99	75	65	50	34	47	34	28
7 ft (2.13 m)												
Lead												
mm	1.7	1.2	0.8	.4	2.4	1.9	1.5	1.0	0.6	1.0	0.6	.3
in.	1/16	1/16	1/32	—	3/32	1/16	1/16	1/32	1/32	1/32	1/32	—
psf†	4.0	2.8	1.9	0.9	6.6	4.4	3.5	2.3	1.4	2.3	1.4	0.7
Concrete												
in.	5.3	3.8	2.8	1.4	7.2	6.1	5.0	3.9	2.8	3.2	2.0	1.1
psf†	65	47	34	17	88	75	61	39	25	39	25	13
10 ft (3.05 m)												
Lead												
mm	1.4	1.0	0.7	.2	2.2	1.7	1.3	0.8	.4	.8	.4	.1
in.	1/16	1/32	1/32	—	3/32	1/16	1/16	1/32	—	1/32	—	—
psf†	3.3	2.3	1.6	0.5	5.1	4.0	3.0	1.9	0.9	1.9	0.9	.2
Concrete												
in.	4.6	3.2	2.2	0.8	6.6	5.3	4.0	2.8	1.4	2.8	1.4	0.4
psf†	56	39	27	10	81	65	53	34	17	34	17	5
14 ft (4.26 m)												
Lead												
mm	1.2	0.8	.5	0	2.0	1.5	1.1	0.6	.2	.6	.2	0
in.	1/16	1/32	—	—	3/32	1/16	1/32	1/32	—	1/32	—	—
psf†	2.8	1.9	1.2	—	4.7	3.5	2.5	1.4	0.5	1.4	0.5	—
Concrete												
in.	3.8	2.8	1.7	0	6.2	5.0	2.6	2.1	0.8	2.1	0.8	0
psf†	47	34	21	—	76	61	43	26	10	26	10	—

*From N.B.S. Handbook 76, 1961, p. 32.

†Pounds per square foot, computed from millimeters of lead or inches of concrete and a concrete density of 2.35 gm/cm³ (147 lbs/ft³).

TABLE 7. SHIELDING REQUIREMENTS FOR BUSY 300 KVP THERAPEUTIC INSTALLATIONS*

(W=40,000 ma-min/week)

Distance from Tube to Occupied Area	Type of Barrier											
	For Controlled Areas						For Environs					
	Primary			Sec- ondary	Primary			Secondary				
UT	1	1/4	1/16	1	1	1/4	1/16	1/64	1/256	1	1/4	1/16
5 ft (1.52 m)												
Lead†												
mm	17.8	14.9	12.1	10.6	22.1	19.2	16.4	13.8	11.2	14.9	12.3	9.7
in.	11/16	19/32	15/32	13/32	7/8	3/4	21/32	17/32	7/16	19/32	1/2	3/8
psf‡	41.3	34.4	27.9	24.6	51.3	44.4	37.8	31.7	25.8	34.6	28.5	22.5
Concrete†												
in.	21.6	19.3	16.8	13.2	25.6	23.3	20.8	18.4	16.0	17.2	14.8	12.4
psf‡	265	237	206	162	314	286	255	226	196	211	181	152
7 ft (2.13 m)												
Lead†												
mm	16.3	13.5	10.7	9.4	20.6	17.8	15.1	12.6	10.0	13.7	11.1	8.5
in.	21/32	17/32	13/32	3/8	13/16	11/16	19/32	1/2	3/8	17/32	7/16	11/32
psf‡	37.8	31.1	24.9	21.9	47.9	41.1	34.9	28.8	22.6	31.8	25.8	19.8
Concrete†												
in.	20.5	18.0	15.7	12.0	24.5	22.0	19.7	17.3	14.9	16.0	13.6	11.2
psf‡	251	221	192	147	300	270	241	212	183	196	167	137
10 ft (3.05 m)												
Lead†												
mm	14.9	12.1	9.4	8.1	19.2	16.4	13.8	11.2	8.6	12.3	9.7	7.1
in.	19/32	15/32	3/8	5/16	3/4	21/32	17/32	7/16	11/32	1/2	3/8	9/32
psf‡	34.6	28.1	22.0	19.1	44.6	38.1	32.0	26.0	20.0	28.8	22.7	16.7
Concrete†												
in.	19.3	16.8	14.4	10.8	23.3	20.8	18.4	16.0	13.6	14.8	12.4	10.0
psf‡	237	206	172	132	285	255	221	191	162	181	152	123
14 ft (4.26 m)												
Lead†												
mm	13.5	10.7	8.3	6.8	17.8	15.1	12.6	10.0	7.4	11.1	8.5	5.9
in.	17/32	7/16	5/16	9/32	23/32	5/8	1/2	13/32	5/16	7/16	11/32	1/4
psf‡	31.3	25.1	19.3	15.8	41.3	35.1	29.2	23.2	17.2	25.8	19.8	13.7
Concrete†												
in.	18.0	15.7	13.2	9.6	22.0	19.7	17.3	14.9	12.4	13.6	11.2	8.8
psf‡	221	192	162	118	270	241	211	181	152	167	137	108
20 ft (6.10 m)												
Lead†												
mm	12.1	9.4	7.0	5.4	16.4	13.8	11.2	8.6	6.1	9.7	7.1	4.5
in.	15/32	3/8	9/32	7/32	21/32	17/32	7/16	11/32	1/4	3/8	9/32	3/16
psf‡	28.1	22.1	16.3	12.6	38.1	32.0	26.2	20.2	14.2	22.5	16.5	10.5
Concrete†												
in.	16.8	14.4	12.0	8.4	20.8	18.4	16.0	13.6	11.2	12.4	10.0	7.6
psf‡	206	176	147	103	255	225	196	167	137	152	123	93

*From N.B.S. Handbook 76, 1961, p. 38.

†Constant potentials may require 15 to 25 percent larger thicknesses of lead and 5 to 15 percent larger thicknesses of concrete than those given here for pulsating potentials.

‡Pounds per square foot computed from millimeters of lead or inches of concrete and a concrete density of 2.35 g/cm³ (147 lb/ft³).

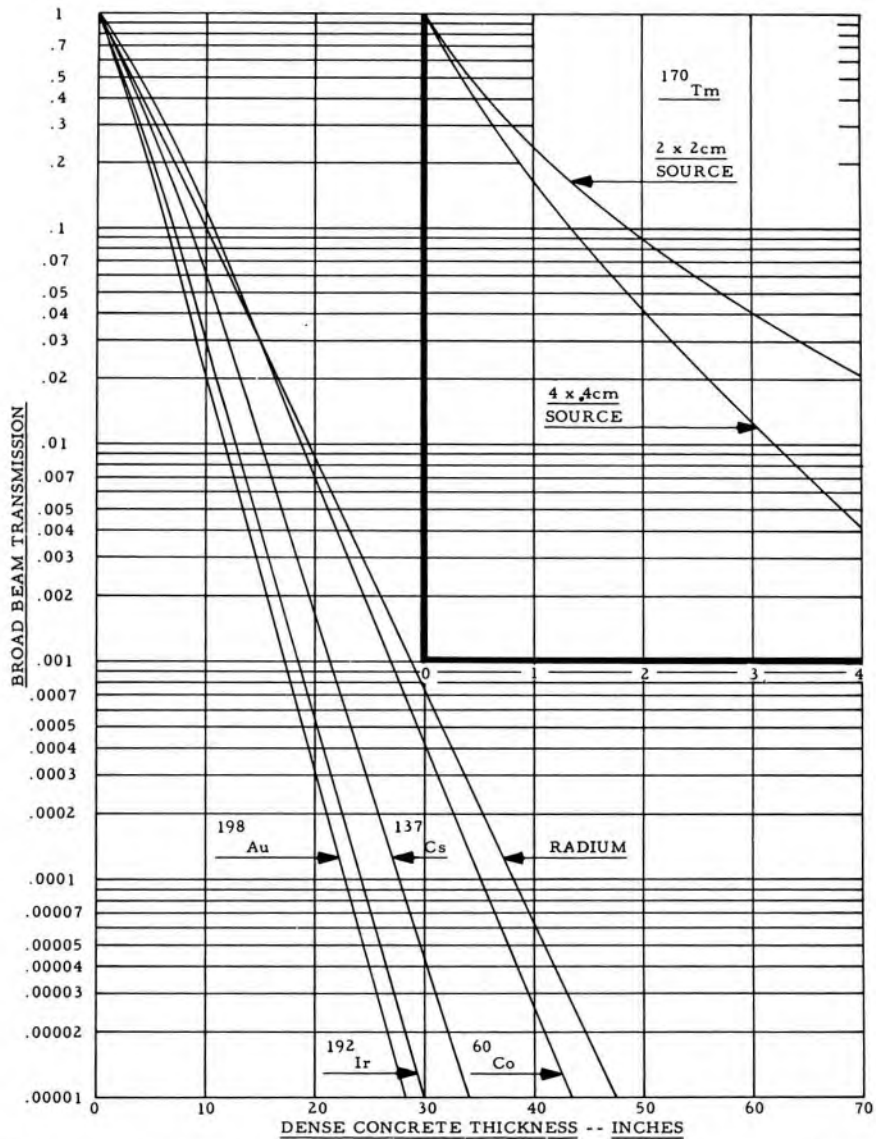


Fig. 3. Broad beam transmission curves of gamma rays from several radionuclides for dense concrete (147 lb/ft^3).¹³

pose. Workload W is taken as the mA-min weekly operation of the machine. These three quantities, in combination with distance, effectively specify both the quality and dosage rate of the unshielded beam.

Three other variables, the type of barrier, UT product, and nature of occupancy, affect barrier calculations. Primary are generally much heavier than secondary barriers, often twice as thick; this is a major consideration in supervoltage installations. The UT product and type of occupancy have been previously discussed.

TELE THERAPY UNITS. Considerations of the UT factor and type of occupancy are the same for teletherapy units as for x-ray machines.

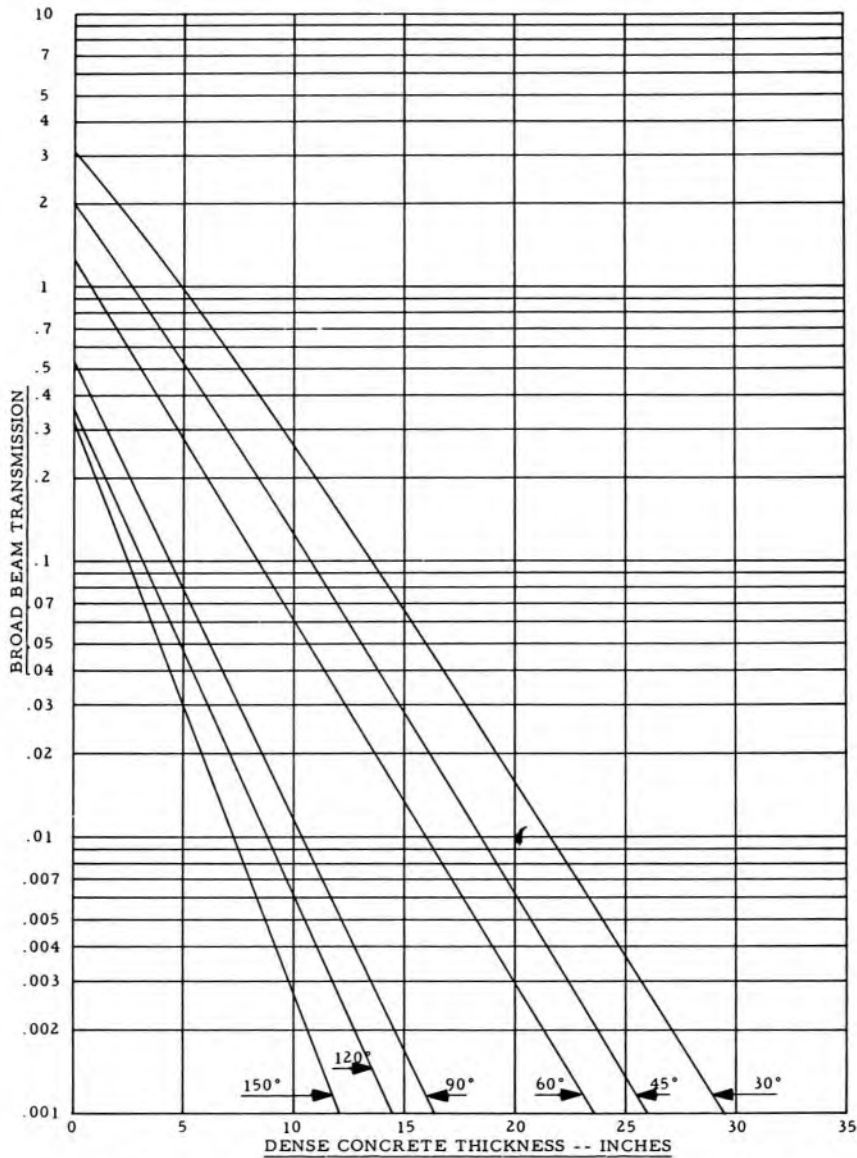


Fig. 4. Attenuation in dense concrete (147 lb/ft³) of cobalt-60 gamma rays scattered from a cylindrical masonite phantom irradiated by a 20 cm diameter field. Ordinate is 1,000 (D_s/D_u), where D_s is the scattered dosage at 1 meter from phantom, and D_u is the air-R dosage at phantom surface.¹⁴

The calculation of unshielded weekly dosage D_o is different, however, as we shall see in the example below. It is computed from the beam intensity, using the inverse square law (equation 12-4).

The basic data used in computing barrier thicknesses are broad beam transmission curves for primary and scattered beams of various radionuclides. Figures 3 and 4 are adapted from N.B.S. H. 73. They give transmission data of primary and scattered beams, respectively. (They are for concrete; similar curves are available for lead, iron, and other materials.) Note in Figure 4 that radiation scattered through small angles is both harder and more intense than that scattered

through greater angles. This fact is of great importance not only in teletherapy but also in supervoltage barrier design.

RADIOACTIVITY STORAGE. Calculations are relatively simple for this application, involving a straightforward application of the approach of Table 1. D_0 is easily computed by (12-4) from the distance, source activity, and Γ . Once the required maximum permitted transmission is known, Figure 5 may be used to obtain the lead thickness required to attenuate gamma rays from various radionuclides. Curves of other radionuclides can be estimated from those in Figure 5, using the relevant decay schemes.

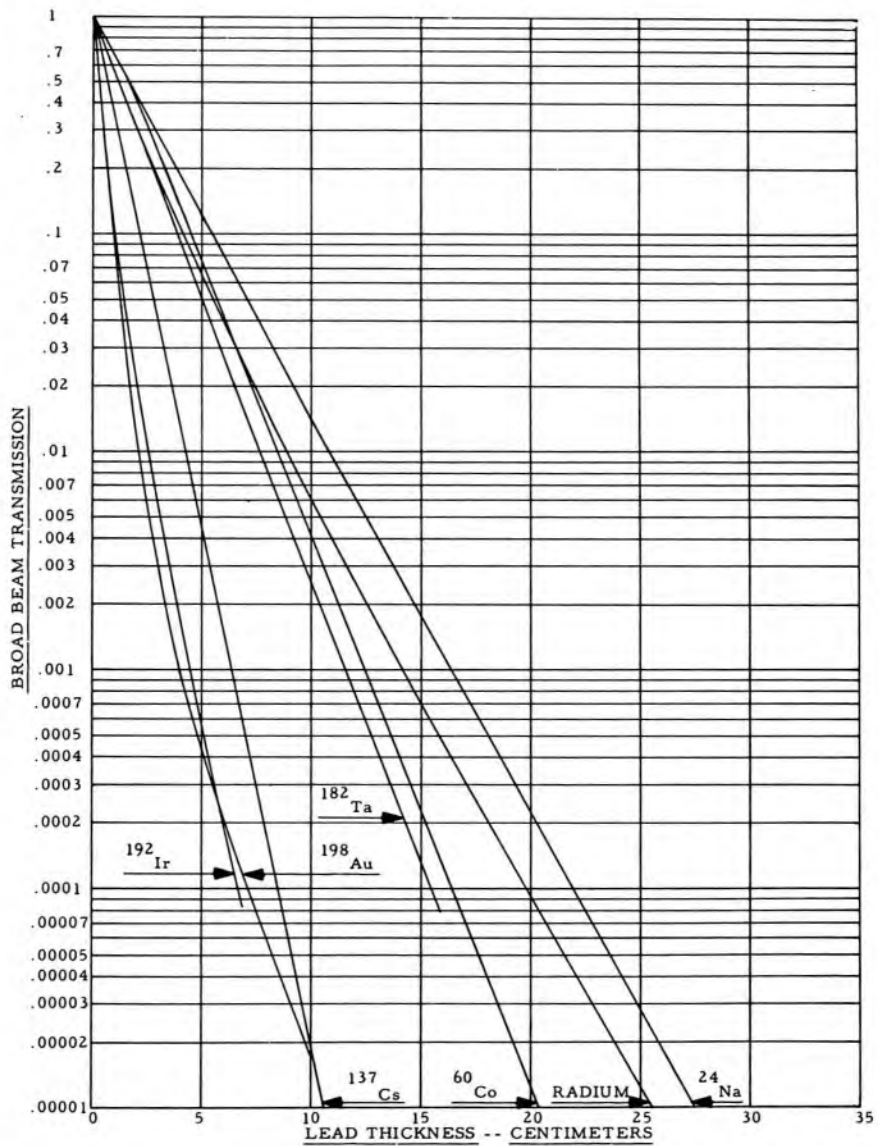


Fig. 5. Broad beam transmission curves of gamma rays from several radionuclides for lead.¹⁵

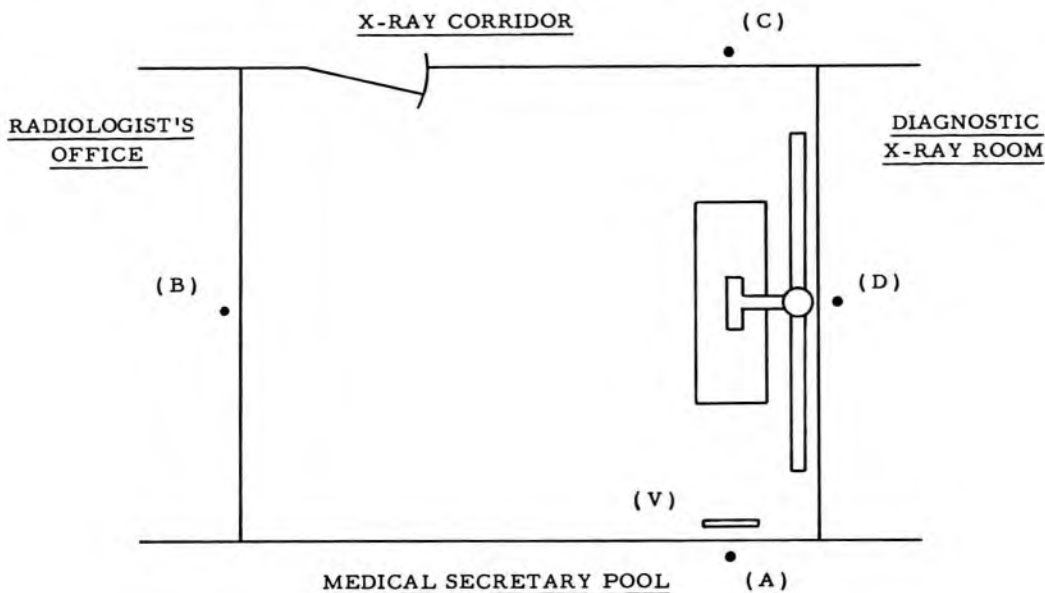
ILLUSTRATIVE EXAMPLES

We shall design protection barriers for four installations. The first two contain a 125 pkV radiographic and a 300 pkV orthovoltage therapy x-ray machine; the third, a cobalt-60 teletherapy unit; and the last, a radium storage area. Installations of high energy super-voltage units (4 to 22 MeV and above) are not included because their barrier designs involve additional considerations beyond this discussion.

Radiographic X-ray Installation

Figure 6 shows a 14×17 foot radiographic room and surrounding areas. In normal use the beam is aimed downward primarily. Chest radiography is also carried out, however, using the vertical cassette holder V on wall A. In addition, the useful beam sometimes strikes parts of walls B and C. Wall D, the ceiling, and walls A, B, and C above 7 feet are all secondary barriers because the beam cannot be aimed at wall D for mechanical reasons, and no useful radiographic study requires aiming the beam much above the horizontal.

DESIGN ASSUMPTIONS. Table 6 is applicable to busy radiographic installations. Note that the one table is applicable to any operating



- NOTES: 1. ROOM ABOVE --- PATHOLOGY LAB; 10 FOOT CEILING
2. ROOM BELOW --- BUSINESS OFFICE

Fig. 6. Plan view of radiographic installation and surrounding areas: 14 x 17 ft room (see text).

kilovoltage from 100 through 150 pkV maximum, which covers all current diagnostic use. We have assumed this to be a busy installation. For the various kilovoltages shown, the mA-min weekly figures of the table are quite adequate, and computed barrier thicknesses will be more than adequate.

We make the following use factor assumptions (Table 8):

Floor	$U = 1$
Wall A near cassette holder	$U = \frac{1}{2}$
Rest of wall A; walls B and C	$U = \frac{1}{4}$

As indicated in the previous section, secondary barriers all have unity use factors, since they receive stray radiation whenever the x-rays are turned on.

Table 8 shows steps in barrier calculations for this installation. Note the rooms above and below as well as the secretarial pool are all considered environs. Some of the people involved may be members of the radiology group. It nevertheless appears prudent to minimize their exposure because they do not employ x-rays as part of their

TABLE 8. RADIOGRAPHY: STEPS IN BARRIER CALCULATIONS OF FIGURE 6 (see text)

A. Primary Barriers

- Floor: distance = 10 ft; $U = 1$; $T = 1$; environs.
(A) $UT = 1$.
(B) Required barrier is 2.2 mm or $\frac{3}{32}$ " Pb, or 6.6" of concrete.
- Wall A: distance = 5 ft; $U = \frac{1}{2}$; $T = 1$; environs.
(A) $UT = \frac{1}{2}$.
(B) Required barrier is less than 2.7 mm or $\frac{3}{32}$ " Pb, or 8.1" of concrete.
- Wall B: distance = 14 ft; $U = \frac{1}{4}$; $T = 1$; controlled area.
(A) $UT = \frac{1}{4}$.
(B) Required barrier is 0.8 mm or $\frac{1}{32}$ " Pb, 1.9" concrete.
- Wall C: distance = 7 ft; $U = \frac{1}{4}$; $T = 1$; controlled area.
(A) $UT = \frac{1}{4}$.
(B) Required barrier is 1.2 mm or $\frac{1}{16}$ " Pb, 2.8" concrete.

B. Secondary Barriers

- Wall D: distance = 5 ft; $U = 1$; $T = 1$; controlled area.
(A) $UT = 1$.
(B) Required barrier is 0.5 mm Pb or 1.7" concrete.
- Ceiling: distance = 7 ft; $U = 1$; $T = 1$; environs.
(A) $UT = 1$.
(B) Required barrier is 0.4 mm Pb or 1.4" concrete.

work and receive no training in radiation safety. The remaining locations are all controlled areas. U and T factors are given in the table.

CONSTRUCTION RECOMMENDATIONS. Barrier thicknesses have been computed for both lead and dense concrete. Since this is a frame building, the added weight of concrete poses support problems, making this material impractical. The recommended lead barriers follow:

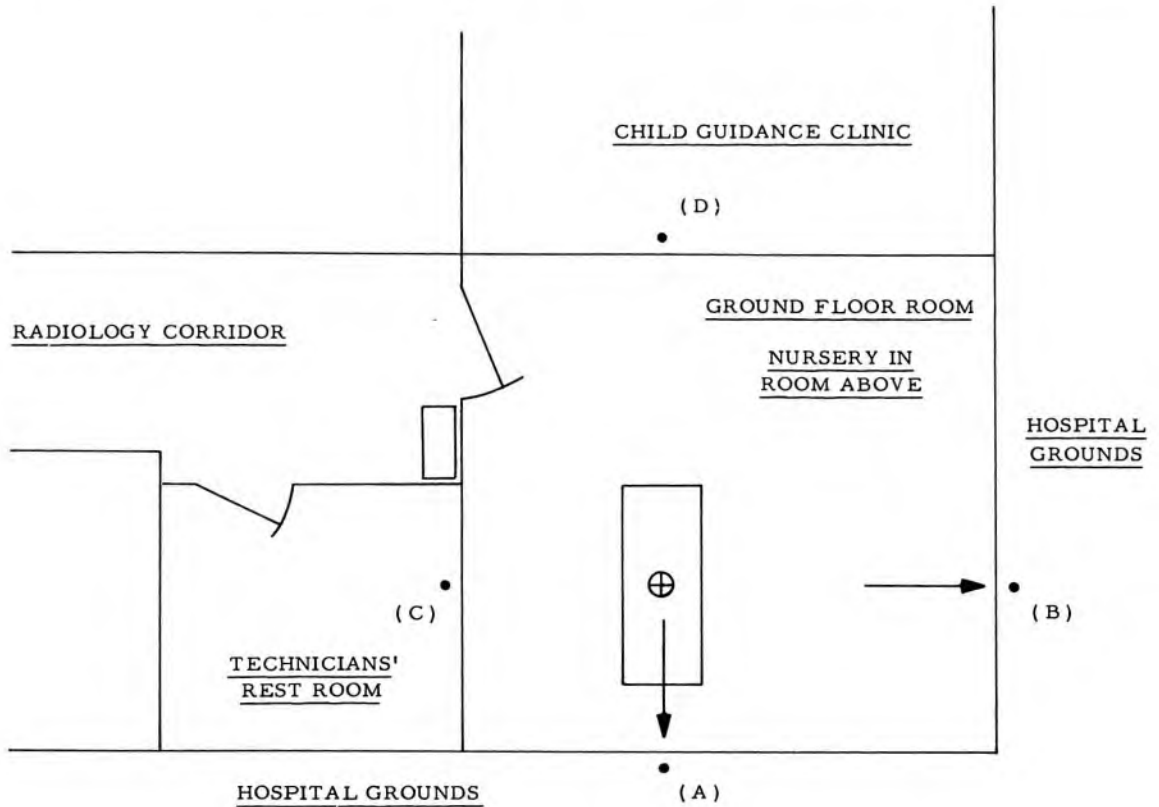
1. *The floor.* Lay a lead mat $3/32$ inch thick to cover the full area of the floor. This can be covered by a wood sub-flooring and suitable floor tile.
2. *The walls.* Some walls require up to $3/32$ inch of lead, others virtually nothing. Standardization of lead thickness is desirable. A $1/16$ inch lead thickness bonded to plywood to 7 foot height may be used throughout, except for the cassette half of wall A, which should be $3/32$ inch. Doors and other openings require the same barrier protection as adjacent walls.
3. *The ceiling.* This is tantalizingly marginal. In a modern office building with concrete floors no added shielding would be required. Even in this building the required 1.4 inch of concrete equivalent may already be present in the ceiling and floor above the machine. A $1/32$ inch lead mat may be placed on the floor above. Alternatively, monthly film badges can be placed beneath chairs, shelves, and desks and given to personnel for a representative period of several months, in hopes readings will demonstrate the mat is unnecessary.

300 pkV Orthovoltage X-ray Installation

As previously shown [Fig. 2(Top)], required lead thicknesses for a given attenuation rise rapidly in the range of 100 to 300 pkV. Consequently, orthovoltage machines are associated with substantially higher lead weight and installation costs than diagnostic machines.

Figure 7 shows a treatment room 15×16 feet in size. The tube, treatment couch, and surrounding areas are also indicated. The beam is ordinarily directed downward but may also be aimed at walls A and B (arrows). The ceiling and walls C and D are all secondary barriers.

DESIGN ASSUMPTIONS. Table 7 is applicable to this installation, from the point of view of kilovoltage and wave form. A machine of this type likely operates at 20 mA. (For example, a Picker *Vanguard* unit might operate at somewhat less than 300 pkV and 20 mA.) For 20



- NOTES: 1. GROUND FLOOR LOCATION
2. NURSERY ON FLOOR ABOVE: 10 FOOT CEILING

Fig. 7. Plan view of 300 pkV orthovoltage radiotherapy x-ray installation and surrounding areas: 15 x 16 ft room (see text).

hours actual operation during the 40 hour week the actual mA-min is:
 $W = (20\text{mA}) (20 \times 60 \text{ min}) = 24,000 \text{ mA-min}$. (This is only half the 40,000 mA-min of the table, so we actually require roughly one HVL less than the thicknesses given in the table.) In practice, the traffic and setup times in even efficient departments prevent x-ray operation much more than 25 to 50 percent of the total time.

We make the following use factor U assumptions:

$$\begin{array}{ll} \text{Floor} & U = 1 \\ \text{Walls A and B} & U = 1/2 \end{array}$$

For secondary barriers, $U = 1$. These are the ceiling and walls C and D.

The room above, child guidance clinic, and outside areas are environs; the others are controlled areas. Occupancy factors are fortunately $T = 1/16$ for the grounds beyond walls A and B (limited access is assumed) but must be taken as $T = 1$ for the controlled areas, the clinic, and nursery.

TABLE 9. 300 pKv ORTHOVOLTAGE THERAPY:
STEPS IN BARRIER CALCULATIONS OF FIGURE 7
(see text)

A. Primary Barriers

1. Floor: no barrier specifically, packed earth beneath a good concrete slab assumed.
2. Wall A: distance = 5 ft; $U = \frac{1}{2}$; $T = \frac{1}{16}$; environs.
(A) $UT = \frac{1}{32}$; our $W = \frac{1}{2}$ that of table, so use $UT = \frac{1}{64}$ in table.
(B) Required barrier is $1\frac{7}{32}$ " Pb, or 18.4" concrete.
3. Wall B: distance = 10 ft; $U = \frac{1}{2}$; $T = \frac{1}{16}$; environs.
(A) $UT = \frac{1}{32}$ ". Use $\frac{1}{64}$ in table because of our lower W value.
(B) Required barrier is $\frac{7}{16}$ " Pb, or 16" concrete.

B. Secondary Barriers

1. Ceiling: distance = 7 ft; $U = 1$; $T = 1$; environs.
(A) $UT = 1$. Use $UT = 1$ in table, for extra factor of 2 in safety.
(B) Required barrier is $1\frac{7}{32}$ " Pb or 16" concrete.
2. Wall C: distance = 5 ft; $U = 1$; $T = 1$; controlled area.
(A) $UT = 1$.
(B) For $UT = 1$, required barrier is $1\frac{3}{32}$ " Pb or 13.2" concrete.
(C) Here, we can drop it an HVL value, to about $1\frac{1}{32}$ " Pb or 12" concrete.
3. Wall D: distance = 10 ft; $U = 1$; $T = 1$; environs.
(A) $UT = 1$. Use $UT = 1$ in table, for extra factor of 2 in safety.
(B) Required barrier is $\frac{1}{2}$ " Pb, 14.8" concrete.

CONSTRUCTION RECOMMENDATIONS. Computations of the required barriers are indicated in Table 9. The actual construction might be as follows:

1. *The floor.* No added protection is needed beyond the usual concrete slab, if the ground is solid and well packed.
2. *Ceiling.* A $1\frac{7}{32}$ inch lead barrier is required, which can be reduced somewhat if the floor above is concrete. For example, it becomes only $\frac{3}{8}$ inch with a 5 inch thick dense concrete floor above. With new construction the floor above can be accommodated to accept a lead mat without much troublesome elevation above surrounding floors. If a lead mat cannot be used a more expensive treatment room ceiling support is required.
3. *Walls A and B.* Since these are outside masonry walls, they offer some useful shielding, provided any windows are also solidly bricked up. Two courses of good quality brickwork are equivalent to about 5.5 inches of dense concrete. This can reduce the required lead thickness to about $\frac{5}{16}$ inch of lead.

4. *Walls C and D.* The existing inside walls contribute negligible x-ray attenuation, so the new barriers must provide full shielding. The use of concrete results in sacrifice of much room volume, so lead barriers should be installed. Since this is a ground floor, lead bonded to cinder blocks appears to be a reasonable construction choice.

Note that relatively heavy shielding is required for the ceiling and wall D, despite the fact they are secondary barriers. This illustrates the fact environs normally increase barrier cost.

Cobalt-60 Teletherapy Installation

Supervoltage x-ray and teletherapy generally involve considerably heavier and more expensive barriers than orthovoltage installations and some rather specialized design problems. The present example illustrates the general approach involved. Optimum design is very important from the point of view of cost and usefulness of the installation, and the services of a QE are usually advisable.

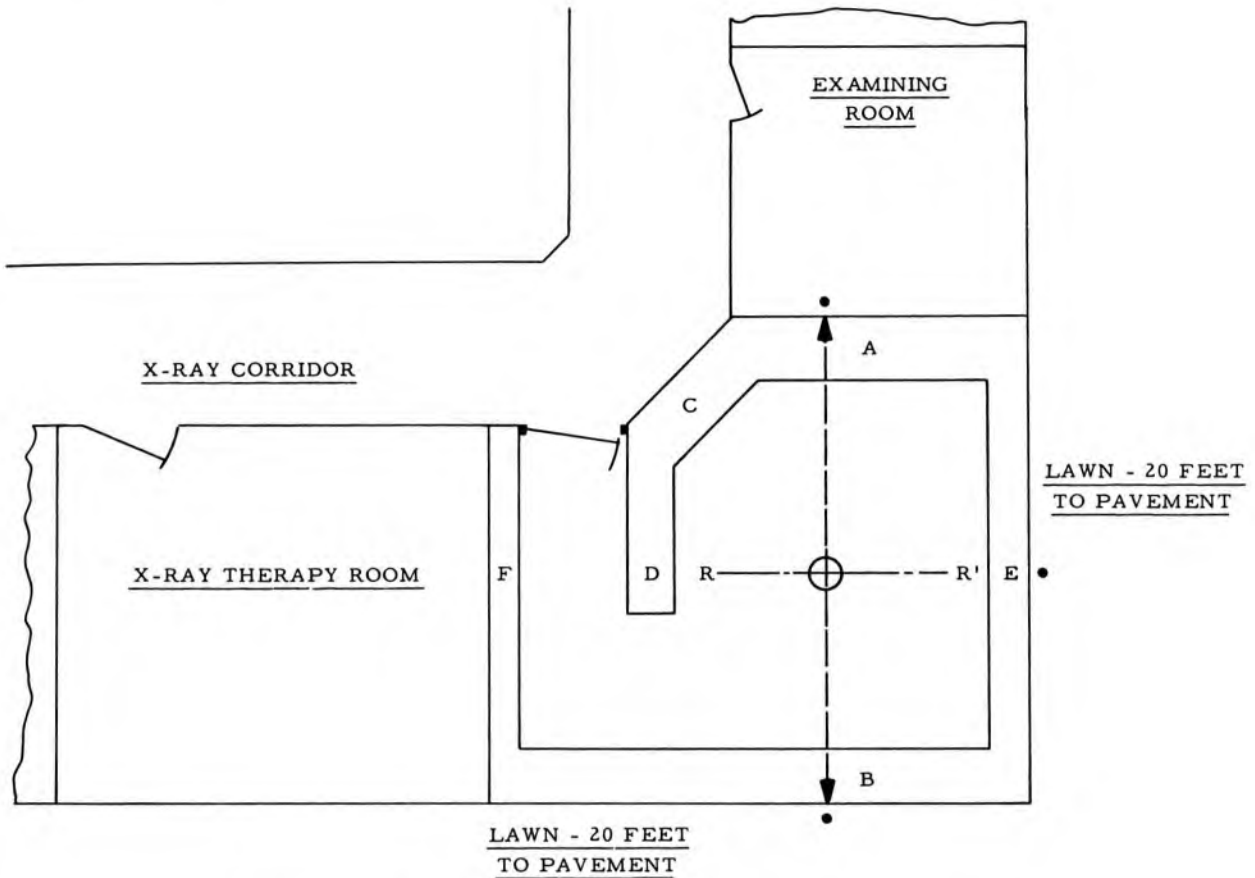
Figure 8 is the plan view of a teletherapy room and its surrounding areas. The room location, on the ground floor with two outside walls, is on the whole quite favorable. These walls face an uncontrolled area; however, the lawn occupancy may be kept low without great difficulty. The other walls face controlled areas. The most serious problem is the ceiling because the room above is full-time occupied environs, and the beam is sometimes directed upward.

In this installation (Fig. 8), the beam central ray is mechanically confined to a vertical plane through the dashed line. The machine is an "isocentric-mount" unit. It may be used for rotational therapy or isocentric multiple portal therapy, with TAD of 80 cm, as well as for the more usual radiotherapy at up to 80 cm SSD. (Rotation of the unit is about the axis RR'.) As a result of this type of operation, the floor, ceiling, and walls A and B are all primary barriers. The rest are by definition secondary, of course.

DESIGN ASSUMPTIONS. Let the source be one of the strongest currently in use, yielding 150 R/m at 1 meter (9,000 R/hr or 9,000,000 mR/hr at 1 meter). Also assume the housing meets NCRP and AEC requirements of 0.1 percent or less leakage when the beam is ON, or 9,000 mR/hr.

Use factors for the primary barriers are as follows:

Floor	$U = 1$
Walls A and B	$U = \frac{1}{2}$
Ceiling	$U = \frac{1}{4}$



- NOTES: 1. GROUND FLOOR LOCATION
 2. ROOM ABOVE IS BUSINESS OFFICE - 12 FOOT CEILING

Fig. 8. Plan view of cobalt-60 teletherapy installation and surrounding areas (see text).

The ceiling is struck only during crossfiring techniques or rotational therapy, so $U = 1/4$ is reasonable. The beam is also directed horizontally for other treatments, so a higher use factor is set for walls A and B than for the ceiling. Of course, the secondary barrier use factors are all unity.

Occupancy is full ($T = 1$) for all the controlled areas as well as for the business office above the treatment room. In accordance with the guidelines of Table 3, the outside lawn areas may be assigned $T = 1/16$; they must be considered environs. The future and even current occupancy of this lawn must be specifically prevented from exceeding this T value, and we shall return to this point below.

Two special AEC licensure requirements must also be met. First, all calculations must assume the beam is turned on for a full 50 hours per week, and second, the beam intensity in all uncontrolled areas should be less than 2 mR/hr, even when occupancy is low. Otherwise special provisions may be required to assure sufficiently low occupancy.

EXIT SHIELD. Unless an exit shield (Chap. 14) is used, very thick wall barriers may be required—up to 30 to 40 inches thick dense concrete and more! In this installation the exit shield is essential because the beam is aimed at the ceiling. Support of a massive ceiling can become an expensive structural problem; in addition, there are only a little more than 3 feet available above the 9 foot ceiling to the floor above, so expensive dense concrete might be required. Hence, an exit shield offers an attractive alternative. However, the shield is clumsy and prevents the use of the beam for whole body or extensive field irradiation unless the housing can be angulated at will.

The exit shield must intercept radiation scattered by the patient through angles of less than $\pm 30^\circ$ (Chap. 14). Photon beams scattered through less than this angle are both harder and more intense than those scattered through greater angles, so the $\pm 30^\circ$ scatter requires a thicker barrier. Using data from N.B.S.H. and book 73, it is interesting to compare the intensity of the direct transmission of a good exit shield with that of the measured 30° , 60° , and 90° scatter, for an 80 cm SSD:

Transmitted beam dose	0.0005 D_0
30° scatter dose	0.0050 D_0
60° scatter dose	0.00203 D_0
90° scatter dose	0.000828 D_0

Thus, for a 1/20 percent transmission barrier, the 30° scatter is 10 times more intense than the direct transmission, and the 90° scatter is only about 1.6 times more intense. Since the 30° scatter is also quite hard, in practice it determines the required barrier thickness.

DATA CHARTS. Basic data are given in Figures 3 and 4 for the transmission of primary and scattered gamma radiation through dense concrete. For our problem the cobalt-60 primary beam transmission data of Figure 3 and the 30° , 60° , and 90° scatter transmission data of Figure 4 are of specific interest.

In this installation, the unattenuated dose rate is 9,000,000 mR/hr, and the various basic dose rates at a meter (for calculation purposes) become:

Transmitted beam dose	4,500 mR/hr
30° scatter dose	45,000 mR/hr
60° scatter dose	18,270 mR/hr
90° scatter dose	7,450 mR/hr

CONSTRUCTION RECOMMENDATIONS. The calculation procedure and required barriers are indicated in Table 10. The actual construction might be as follows:

TABLE 10. COBALT-60 TELETHERAPY:
CALCULATIONS FOR FIGURE 8

A. Primary Barriers (see Figures 3 and 4)

1. Floor: no special barrier.

2. Wall A: distance = 10 ft; $D^2 = 10 \text{ m}^2$; $U = 1/2$; $T = 1$; controlled area.

(A) Through beam, with exit shield

(1) $Do = [4,500 \times 50][1/2 \times 1][1/10] \text{ mR/wk} = 11,250 \text{ mR/wk}$.(2) Reduction factor = $100/11,250 = .0089$ —18" concrete (Fig. 3).

(B) 30° scatter

(1) $Do = [45,000 \times 50][1/2 \times 1][1/10] = 112,500 \text{ mR/wk}$ (2) Reduction factor = $100/112,500 = .00089$.(3) Required curve ordinate = $3.2 (.00089) = .00285$ —26" concrete (Fig. 4).(4) This beam is 30° oblique to wall, so only $\sqrt{3}/2$ as much is needed, or 23" concrete.

(C) Hence, use 24" concrete.

(D) Beam intensity to area = $(100 \text{ mR}/50 \text{ hrs}) \times (2 \times 1) = 4 \text{ mR/hr}$.3. Wall B: distance = 9 ft, $D^2 = 8 \text{ m}^2$; $U = 1/2$; $T = 1/16$; environs.

(A) Through beam, with exit shield

(1) $Do = [4,500 \times 50][1/2 \times 1/16][1/8] = 880 \text{ mR/wk}$.(2) Reduction factor = $10/880 = .0113$ —18" concrete (Fig. 3).

(B) 30° scatter

(1) $Do = [45,000 \times 50][1/2 \times 1/16][1/8] = 8,800 \text{ mR/wk}$ (Fig. 10).(2) Reduction factor = $10/8,800 = 1/880 = .00113$.(3) Required curve ordinate = $3.2 (.00113) = .0036$ —25" concrete (Fig. 10).

(4) Obliquity reduces this to 22" concrete.

(C) Use 24" concrete.

(D) Beam intensity to area = $\frac{10 \text{ mR}}{50 \text{ hrs}} \times (2 \times 16) = 6.4 \text{ mR/hr}$.4. Ceiling: distance = 12 ft, $D^2 = 14.3 \text{ m}^2$; $U = 1/4$; $T = 1$; environs.

(A) Through beam, with exit shield

(1) $Do = (4,500 \times 50)(1/4 \times 1)(1/14.3) = 3,940 \text{ mR/wk}$.(2) Reduction factor = $10/3,940 = 1/394 = .00254$ —23" concrete (Fig. 3).

(B) 30° scatter

(1) $Do = (45,000 \times 50)(1/4 \times 1)(1/14.3) = 39,400 \text{ mR/wk}$.(2) Reduction factor = $10/39,400 = 1/3,940 = .000254$ (3) Required 30° curve ordinate = $3.2 (.000254) = .00081$ —30" concrete (Fig. 4).

(4) Obliquity reduces this to 26" concrete.

(C) Use 28" concrete.

(D) Intensity = $\frac{10 \text{ mR}}{50 \text{ hrs}} (4 \times 1) = 0.8 \text{ mR/hr}$.

B. Secondary Barriers (Fig. 4)

1. Wall C: This barrier is struck at one end by the 30° scatter from the beam aimed at wall A. Since wall C is short, make it 26" concrete.

(continued)

TABLE 10. COBALT-60 TELETHERAPY: CALCULATIONS
FOR FIGURE 8 (continued)

2. Wall D: distance = 10 ft, $D^2 = 10 \text{ m}^2$; $U = \frac{1}{2}$; $T = \frac{1}{4}$; controlled area.
 - (A) This barrier shields the door area and corridor, mainly, from 60° scatter from patient when beam is aimed at wall A.
 - (B) $Do = (18,270 \times 50)(\frac{1}{2} \times \frac{1}{4})(\frac{1}{10}) = 11,400 \text{ mR/wk}$.
 - (C) Reduction factor = $\frac{100}{11,400} = .0087$.
 - (D) Required 60° curve ordinate = $1.3 (.0087) = .0113-16''$ concrete.
 - (E) Obliquity reduces this to $14''$ concrete.

3. Wall E: distance = 8 ft, $D^2 = 6.33 \text{ m}^2$; $U = 1$; $T = \frac{1}{16}$; environs.
 - (A) 90° scatter only.
 - (B) $Do = (7,450 \times 50)(1 \times \frac{1}{16})(\frac{1}{6.33}) = 3,680 \text{ mR/wk}$.
 - (C) Reduction factor = $\frac{10}{3,680} = .0027$.
 - (D) Required ordinate = $0.53 (.0027) = .0014-16''$ concrete.
 - (E) Use $16''$ concrete.
 - (F) Beam intensity = $\frac{10 \text{ mR}}{50 \text{ hrs}} (16) = 3.2 \text{ mR/hr}$.

4. Wall F: distance = 13 ft, $D^2 = 16.7 \text{ m}^2$; $U = 1$; $T = 1$; controlled area.
 - (A) 90° scatter only.
 - (B) $Do = (7,450 \times 50)(1 \times 1)(\frac{1}{16.7}) = 22,300 \text{ mR/wk}$.
 - (C) Reduction factor = $\frac{10}{22,300} = .00045$.
 - (D) Required 90° curve ordinate = $0.53 (.00045) = .00024-20''$ concrete.

1. *The floor.* The ground must be well packed and solid to assure proper self-absorption of radiation scattered upward near the walls from below floor level. A 1 foot thick slab of dense reinforced concrete is also recommended.
2. *Wall A.* At least 24 inches of dense concrete. The beam intensity through a 24 inch barrier is more than 2 mR/hr, so special warning signs may be required by the AEC. An increase to 28 inches is desirable to avoid this requirement in the examining room.
3. *Wall B.* At least 22 inches of dense concrete. Here too the beam intensity is a bit high, and the use of 24 inches would be recommended. The lawn should also be enclosed with a locked fence and warning signs provided to assure limited access.
4. *Wall C.* At least 26 inches of dense concrete. Actually, the 30° scatter beam strikes this wall less obliquely than wall A, so more concrete is recommended. This could be tapered down to 20 inches near wall D, but a uniform thicker barrier may actually be cheaper in practice.
5. *Wall D.* At least 20 inches of dense concrete.
6. *Wall E.* At least 16 inches of dense concrete. The 3.2 mR/hr beam intensity is a bit high, and a $T = 1/16$ has been as-

sumed. Therefore, use 18 inches of concrete to be sure. This lawn area should also be fenced off and warning signs provided.

- 7. *Wall F.* At least 14 inches of concrete.
- 8. *Ceiling.* At least 28 inches of concrete. Probably 30 inches is preferable to provide an added factor of safety.

Radium Storage Area

As indicated previously, radium usage involves many hazards besides those of storage. Safe storage is a general problem for fractional curie quantities of several gamma-emitting radionuclides, both sealed and unsealed, so we shall calculate required shielding for a 500 mCi radium supply.

Figure 9 shows the radium location with the type of occupancy in surrounding areas. All barriers are struck by the primary beam for the full time radium is stored, since radioactivity cannot be turned off.

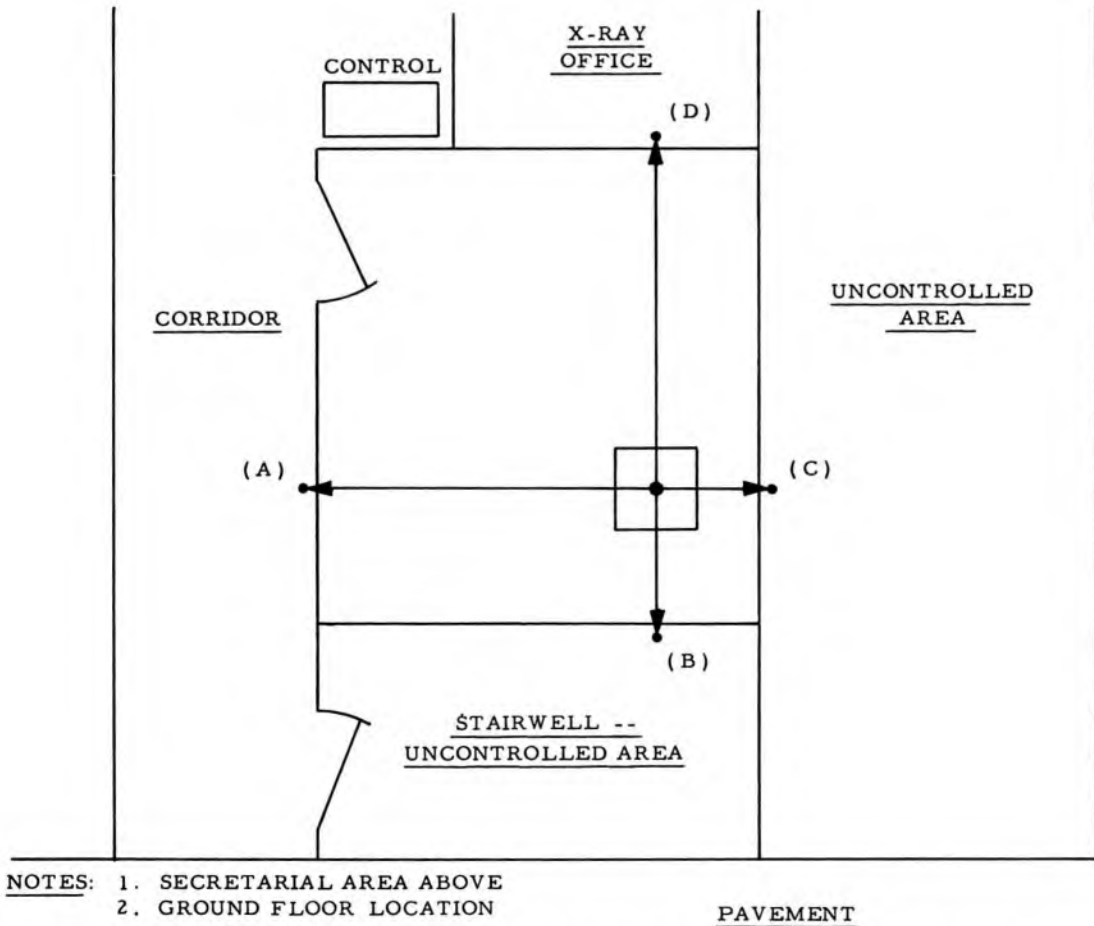


Fig. 9. Plan view of radium storage location and surrounding areas (see text).

DESIGN ASSUMPTIONS. The radium is stored on the ground floor with controlled areas on two sides. Uncontrolled areas exist in the room above and on the two other sides.

As in computing cobalt-60 teletherapy barriers, we must know the unshielded intensity of the source in mR/hr at a meter. The sources may be considered to have negligible self absorption in this problem, so we can employ equation (12-4) to yield:

$$I = \Gamma \frac{M}{d^2} = 8,250 \frac{500}{(100)^2} = 412 \text{ mR/hr at a meter}$$

(This assumes 0.5mm Pt filtration.) The barriers include the safe and any associated lead ell or lead bricks used for shielding.

Table 11 indicates the barrier calculations, with assumed T values; U is unity throughout, as indicated above. Values of the required lead thickness have been obtained using data from Figure 5 for transmission of radium gamma rays through lead.

PRACTICAL ASPECTS. Commercial safes usually provide 2½ inches of lead shielding in all directions. This is evidently adequate for directions towards A and D, occupationally occupied areas. Additional lead is needed to protect the other areas, however. A single additional layer of 2 inch thick lead bricks could be used to protect locations B

TABLE 11. RADIUM STORAGE: SHIELDING CALCULATIONS FOR FIGURE 9

1. Floor: no shielding.
2. Location A: distance = 10 ft = 3.18 m; U = 1; T = 1; controlled.
 - (A) Unshielded weekly dose = $[412 \times 40 \times 1 \times 1] \left[\frac{1}{3.18} \right]^2 = 1,600 \text{ mR}$.
 - (B) Required attenuation = $\frac{10}{1,600} = \frac{1}{16} = .0625$.
 - (C) Needed lead = 6 cm = 2.4 inches of lead.
3. Location B: distance = 4 ft = 1.31 m; U = 1; T = $\frac{1}{16}$; uncontrolled.
 - (A) Unshielded weekly dose = $[412 \times 40 \times 1 \times \frac{1}{16}] \left[\frac{1}{1.31} \right]^2 = 600 \text{ mR}$.
 - (B) Required attenuation = $\frac{10}{600} = .017$.
 - (C) Needed lead = 8 cm = 3.15 inches of lead.
4. Location C: distance = 3 ft or 1 m; U = 1; T = 1; uncontrolled.
 - (A) Unshielded weekly dose = $[412 \times 40 \times 1 \times 1] \left[\frac{1}{1} \right]^2 = 16,000 \text{ mR}$.
 - (B) Required attenuation = $\frac{10}{16,000} = .00063$.
 - (C) Needed lead = 16 cm = 6.3 inches of lead.
5. Location D: distance = 10 ft = 3.18 m; U = 1; T = 1; uncontrolled. This is similar to 2. above.
6. Next floor: distance = 6 ft or 2 m; U = 1; T = 1; uncontrolled.
 - (A) Unshielded weekly dose = $[412 \times 40 \times 1 \times 1] \left[\frac{1}{2} \right]^2 = 4,000 \text{ mR}$.
 - (B) Required attenuation = $\frac{10}{4,000} = .0025$.
 - (C) Needed lead = 12 cm = 4.7 inches of lead.

and the floor above, but *two* layers of bricks are needed to protect location C. It is evident that commercial safes may be inadequate for larger activity radium storage near uncontrolled areas.

People may work inside the room, which might also be used for 250 kV x-ray therapy, for example. For such people as well as those preparing radium, additional protection may be needed, as previously discussed.

It is sometimes suggested that x-ray room shielding protects surrounding areas from radium rays as well. A glance at Table 11 indicates *inches*, not *millimeters* of lead thicknesses are needed. Hence, ordinary diagnostic shielding ($3/32$ inch of lead) helps negligibly, and even orthovoltage shielding ($1/4$ to $5/8$ inch of lead) is only a bit better. Supervoltage and teletherapy beams, however, are more comparable in penetration to radium rays. Consequently, the walls of supervoltage therapy rooms incidentally also shield surrounding areas from radiation from any stored radium.

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16

Miscellaneous: Particle Beam Sources, Diagnostic Ultrasound, and Thermography

Previous chapters have considered the production, properties, and uses of x-rays and radionuclides. Basic x-ray circuits and tubes of conventional machines were discussed in Chapters 2 and 3. However, several important machine designs were necessarily postponed to this chapter; these include some lower kilovoltage applications (20 to 300 kV) and supervoltage x-ray generators. High energy ion beam generators and nuclear reactors have important clinical uses, so their basic principles will also be discussed briefly.

We shall consider in addition two promising diagnostic techniques which use *nonionizing* radiation: ultrasonography and thermography. The former employs megahertz frequency sound waves to evaluate structure and motion of body parts. The latter records the distribution of skin temperature over suspect areas of a patient. Atypical temperature patterns sometimes provide useful clues regarding subcutaneous pathology.

PARTICLE BEAM SOURCES

Photon or x-ray beams are of greatest clinical importance at present. Beams of electrons, ions, and neutrons have been employed for radiotherapy, however, as well as for production of radionuclides, so we shall consider the generation of all four types of particle beams. Photon sources are discussed first, then those of electrons, ions, and neutrons, in order.

Photon Sources

Medical x-ray generators provide an extensive range of photon energies, from 2 keV through 42 MeV; a summary of types is shown

in Table 1. As previously shown (Chapter 3), all x-rays are produced using beams of fast electrons. The operating principles of high energy electron beam accelerators are discussed below. This section deals primarily with other interesting practical aspects of photon generators, both x-ray and teletherapy types.

LOW ENERGY SOURCES. Diagnostic machines employ rotating anode tubes, using previously discussed electrical circuits. Tubes of special design have been employed in high contrast low kV studies (down to 20 pkV) for breast and extremity work.¹ Such tubes have three special requirements. First, their filaments must be mounted relatively close to their anodes. Otherwise excessively high filament temperatures are needed to obtain high tube current, resulting in short tube life; in addition, mA values vary greatly when kV settings are changed.

TABLE 1. PHOTON SOURCES USED IN RADIOLOGY

A. Low Energy

1. Diagnostic

- (A) Ordinary units: 40 to 150 pkV.
- (B) Low kV units: as low as 20 pkV.

2. Therapy

- (A) Ordinary: 60–150 pkV. Moderate filtration.
- (B) Low kV units, with low filtration.
 - (1) Beryllium window tubes; Grenz x-rays, 6 to 15 pkV; and others, to 60 pkV.
 - (2) Chaoul: 60 pkV.
 - (3) Philips: 44 pkV.

B. Orthovoltage Therapy: 140 to over 300 pkV

- 1. Ordinary: Villard, constant potential and self-rectified; 60 Hz supply.
- 2. Newer designs
 - (A) Tube head: Maxitron and Vanguard units; 1,200 Hz.
 - (B) Power supplies: kV stabilizers; selenium rectifiers.

C. Supervoltage Therapy: 1 MV through 42 MV

- 1. Resonant transformer: 1 MV and 2 MV peak.
- 2. Van de Graaff: 1 MV and 2 MV constant potential.
- 3. Linear accelerator: 4 to 8 MV; electrons also. Very short, repetitive pulses.
- 4. Betatron: 9 to 42 MV; electrons also. Very short, repetitive pulses.

D. Teletherapy

- 1. Cesium-137: 0.662 MeV initial photon energy.
- 2. Cobalt-60: 1.17 and 1.33 MeV initial photon energy.

Second, required mAS values are greater for low kV beams because of their low penetration, so anode capacity must be quite great. Finally, inherent filtration is preferably somewhat less than standard. Beryllium window tubes, however, are not required since at least 0.5 mm Al inherent filtration is needed to protect patients' skin.

Most superficial x-ray therapy is carried out using 60–150 pkV, moderate total filtration, and treatment distances of the order of 20 to 30 cm SSD. Therapists, however, also treat with beams generated at 6 to 60 pkV, with very low filtration. For very shallow lesions beryllium window tubes are sometimes employed. X-ray beams from beryllium window tubes are quite intense, constituting a serious hazard in use. Both Chaoul and Philips beams have somewhat greater filtration; they are positioned close to the skin, in so-called "contact therapy." Since these machines are used at short distances to obtain rapid falloff with depth, their field sizes are also rather small. Beryllium window tubes have the advantage of providing much softer x-rays. The very low HVL of their radiation assures very rapid falloff of intensity with depth even at longer SSD values, permitting convenient treatment of larger size portals than contact therapy units. (See Chapter 7.)

ORTHOVOLTAGE—140 TO OVER 300 PKV. We have already discussed conventional 60 cycle units provided with thick window tubes in large, relatively clumsy housings.

Somewhat newer designs have made possible smaller tube head assemblies. These permit more precise and versatile deep therapy around body areas of high curvature, such as the head and neck. Such units employ high insulation, self-rectified tubes.² Housing miniaturization is facilitated by the use of higher frequency voltage, provided by special power oscillators or motor generators. These units supply about 1,200 Hz ac to the high tension transformer. The latter is small enough to fit inside the tube housing; it is usually insulated with gas instead of oil (freon under 3 atmospheres pressure, for example). With both types of 1,200 cycle generators the patient hears a whine during operation, but this is not usually objectionable.

Other technological improvements have also been introduced. For example, selenium rectifiers are more reliable and compact than valve tubes. Also, at least one therapy unit employs an automatic line voltage stabilizer to assure pkV and mA reproducibility.

In recent years, the clinical demand for all orthovoltage machines has declined rapidly, reflecting the increasing preference of radiotherapists for supervoltage radiation.

SUPERVOLTAGE—1 MV THROUGH 42 MV. Resonant transformer and Van de Graaff machines can both produce 1 and 2 MV x-rays and/or

electrons. The resonant transformer units are dependable, stable machines with an extensive record of performance, particularly in industry. Their x-ray outputs are quite high because tube currents are substantial.

Van de Graaff units have slightly lower but quite useful outputs (75 R/m at 1 meter). They are much smaller, fitting into more modest treatment rooms. Their focal spot sizes are smaller than those of resonant transformer units (0.3 versus more than 1 cm), assuring better penumbra characteristics. Although older units initially presented some service problems, the improvement of belts and control circuits and the introduction of sealed tubes have made medical Van de Graaff machines quite reliable. Finally, they are less expensive than most other supervoltage machines. Price and size considerations have apparently resulted in a greater medical application of Van de Graaff than resonant transformer units.

Linear accelerators produce photons in an energy range many radiotherapists consider optimum: 4–8 MeV maximum. In addition their penumbral characteristics are excellent. Output values at 6 MeV are adjustable from 100 to 400 R/m at a meter SSD, with good stability; such high output makes possible a large treatment load. Newer units can rotate about a fixed center ("isocenter"). This is useful for both rotational and multiple port therapy. The required room height is as little as 8 feet when ingenious electron beam-bending and short tube systems are used. Adjustable collimators are provided, and shielding of areas in the treatment field is relatively easily carried out.

The betatron is a very high energy device. Its penumbra and lateral scatter characteristics are excellent, but it has certain practical disadvantages. The lower energy units (22 MeV and less) have relatively low output rates at larger field sizes, compared with other supervoltage units. In addition, collimation presents practical problems because greater lead thicknesses are required to stop the very energetic photons. The larger machines are somewhat heavy and clumsy to position and not movable about the patient. Finally, some of the higher energy units (above 24 MeV) have apparently presented considerable service difficulties up to the time of this writing.

A major advantage of betatrons has been the availability of electron beams, with energies up to 25 MeV in one unit, 35 MeV in another, and 42 MeV in a third. (Newer design clinical linear accelerators are apparently in prospect with capability of electron beams exceeding 30 MeV in energy.) Such electrons can be used in deep therapy. Their application entails many practical and scientific problems, currently the subject of active research. Accurate dosimetry and protection work are required for good results and safety. Electrons are

readily scattered at density discontinuities, so uniform tissue dosage distribution is difficult to obtain in the thorax and other areas of tissue inhomogeneity.³

TELETHERAPY. Both cesium-137 and cobalt-60 have been employed in teletherapy machines.

Cesium-137 is relatively inexpensive to produce since it is a by-product of nuclear reactor operation. In addition, its half-life is quite great so that sources can be used a long time without replacement. Consequently, its use has been intensively studied and many machines constructed. Unfortunately, the gamma constant of cesium-137 is relatively small, and its specific activity only moderate. As a result, intense sources are necessarily very large, with excessive penumbra values. Cesium-137 units are used currently only at short SSD values, primarily for head and neck work.

Cobalt-60 teletherapy machines have become the most common supervoltage modality. They provide the advantage over x-ray units of great dependability, because the photon source need not be serviced and maintained, only replaced about once every five years. Cobalt-60 units are distributed all over the world and are widespread in the United States. As previously indicated, high capacity units are preferable to small ones, for both depth dose and penumbra reasons.

One potential problem, however, is inherent in all teletherapy machines: the shutter can on rare occasions remain fixed in the open position after the treatment is over. Two types of shutters are most commonly used: the rotating wheel and moving block (Fig. 1). In general, instructions for emergency manual closing of these shutters are provided by the manufacturers. The U.S. AEC requires that an

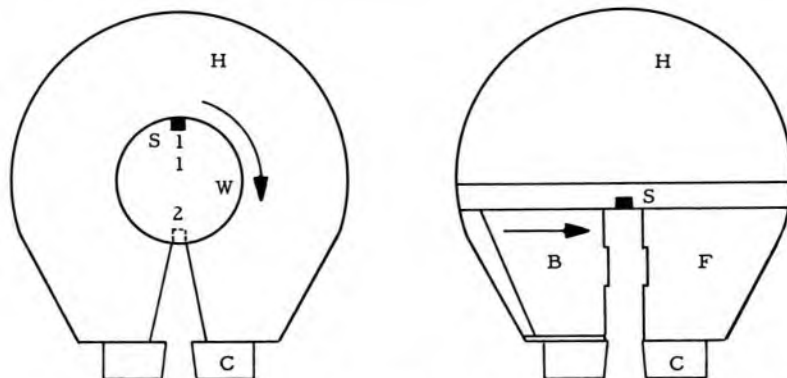


Fig. 1. Two types of teletherapy shutter systems. Left. Wheel type. Wheel W in housing H contains source S. Wheel is made of shielding material, so source is shielded in all positions but 2. Field is defined by collimator C. Right. Moving block type. Block B moves as indicated by arrow, to absorb rays from source in closed position. Mercury shutters have been employed in the past but are not often used nowadays.

emergency procedure be prominently displayed for such an emergency (Chapter 14).

Charged Particle Acceleration

Beams of both electrons and ions have radiologic applications, but those of electrons are so far most important medically. In addition to producing x-rays in tubes, electron beams are also used directly to irradiate tumors, with particle energies of 3 to 42 MeV.⁴ The most commonly used ion beams consist of protons and deuterons accelerated in special machines. Their main medical use is to produce radioisotopes which are either not readily produced in nuclear reactors or of such short half-life that typical shipping delays result in excessive decay before use. Ion beams have also been used for experimental irradiation of certain brain and pituitary lesions.⁵ They offer advantages of minimum lateral scatter and accurately adjustable penetration in soft tissue. These characteristics facilitate delivery of well-localized dosage. An additional advantage of such beams is their high LET values, which imply a reduced oxygen effect.

BASIC REQUIREMENTS OF PARTICLE ACCELERATORS. Specific designs of electron and ion accelerators will be considered separately below. It is helpful, however, to first consider five basic requirements of *all* particle accelerators, indicated in Table 2.⁶ These will now be considered in order.

Electrons are usually supplied by heated filaments. Ion sources generally are more complex, involving gas discharge systems which require special means to inject ions into the tube.

A very good vacuum is essential in any accelerator, as previously described, to prevent electrical field distortion and tube damage from uncontrolled ionization of gas or vapor.

All accelerators employ an electric field to speed up charged particles. The Van de Graaff uses constant potential dc. Both resonant transformer and betatron machines are essentially elegant step-up transformers, employing ac. The cyclotron uses an ingenious electrode arrangement in which ac voltage repetitively accelerates ions traveling in a circular path to high energies. Finally, linear accelerators employ the ac electric fields of radio waves traveling in evacuated cavities to accelerate charged particles.

Of course, the charged particles must be properly confined to the desired path or the system will not function. Electric field focusing and guiding arrangements are employed in x-ray tubes. Magnetic fields are used to both confine electrons and generate the accelerating voltage in a betatron; in other machines constant or adjustable dc mag-

TABLE 2. FIVE BASIC REQUIREMENTS OF CHARGED PARTICLE ACCELERATORS⁶

1. Source of charged particles
 - (A) Electron or ion supply.
 - (B) Insertion or "injection" means.
2. Vacuum
3. Accelerating Voltage
 - (A) Constant dc: Van de Graaff.
 - (B) ac: resonant transformer, betatron, cyclotron.
 - (C) Radio-frequency resonant cavity: linear accelerator.
4. Particle guiding system
 - (A) Electrical and magnetic electron or ion focusing system.
 - (B) Magnetic confining and bending system.
5. Means to use the beam
 - (A) To make x-rays: reflection and transmission targets.
 - (B) To extract particles as beams
 - (1) Straight systems: simple windows of Be or thin Al.
 - (2) Betatrons, cyclotrons, and synchrotrons: extraction coil and magnetic shield system, with window.

netic fields can be used for guiding, focusing, and deflecting electron and ion beams.

We have already discussed the use of targets to produce bremsstrahlung x-rays. In ordinary tubes x-rays emerge from the same surface struck by the electron beam [Fig. 2(Top)]; one generally says such a tube has a "reflection target." Supervoltage x-rays generally emerge from the side of the target *opposite* that struck by electrons; targets are then called "transmission" targets [Fig. 2(Bottom)].

High energy particle beams readily penetrate thin beryllium or aluminum windows. When the beam travels in a straight line no special extraction means is required. In some systems, however, the beam travels in a circular path (i.e., betatron and cyclotron); for these machines, a special magnetic field is usually required to deflect the beam from its normal path towards the window, at just the right time. Pulse circuitry can be designed to do this when the proper beam energy has been achieved.

Medical Electron Accelerators

Four basic types of electron generators are currently of most medical interest: the resonant transformer, Van de Graaff, betatron,

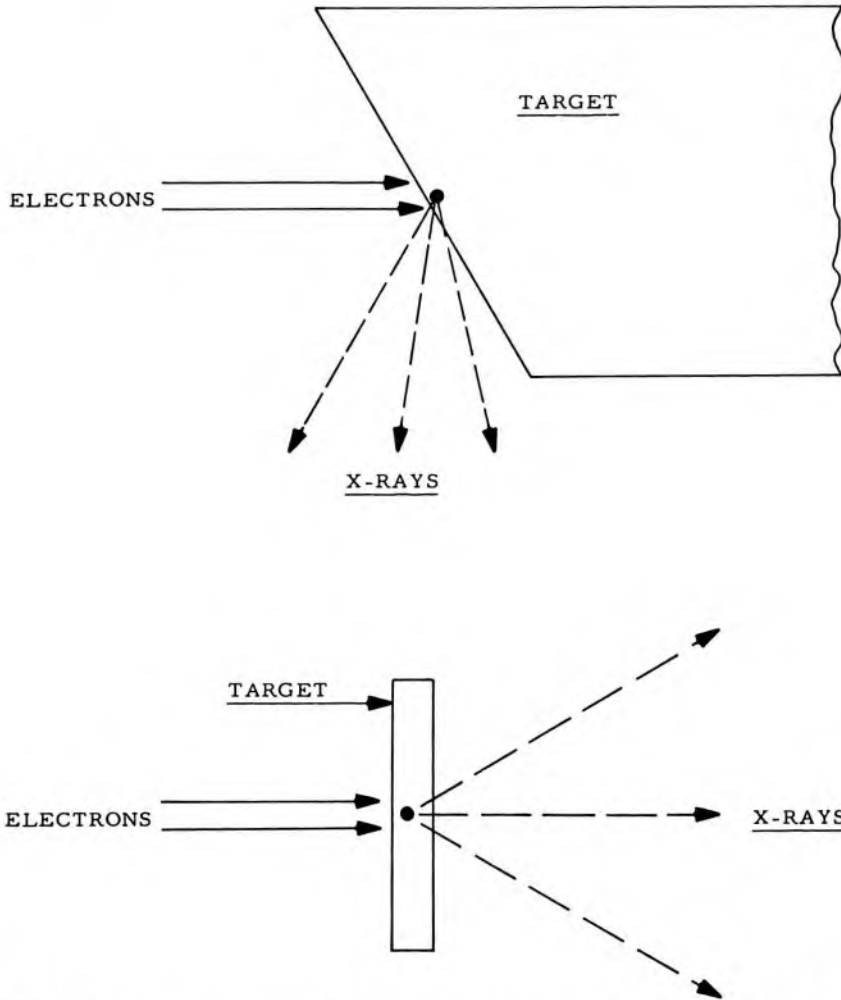


Fig. 2. Reflection vs. transmission x-ray targets. Top. Reflection type. X-rays emerge from same surface struck by electron beam (below 500 pkV). Bottom. Transmission type. X-rays emerge from opposite surface to that struck by electron beam (1 MV and higher, generally).

and linear accelerator machines. The first two units are used primarily below 2 or 3 MeV; the other two, for energies 4 MeV and higher. We shall now consider these machines in order.

RESONANT TRANSFORMER UNIT.⁷ Figure 3 shows a very much simplified schematic diagram of the General Electric Corporation resonant transformer unit. It is essentially a self-rectified machine with grounded anode x-ray tube. The use of very high voltage with substantial tube current, however, necessitates certain special designs:

1. *Tube.* This is constructed in sections, with the applied high voltage divided equally along its length to prevent applying excessive voltage across any particular tube insulator. The transformer secondary winding is correspondingly divided with appropriate staggered connections, or "taps," at appropriate intervals.

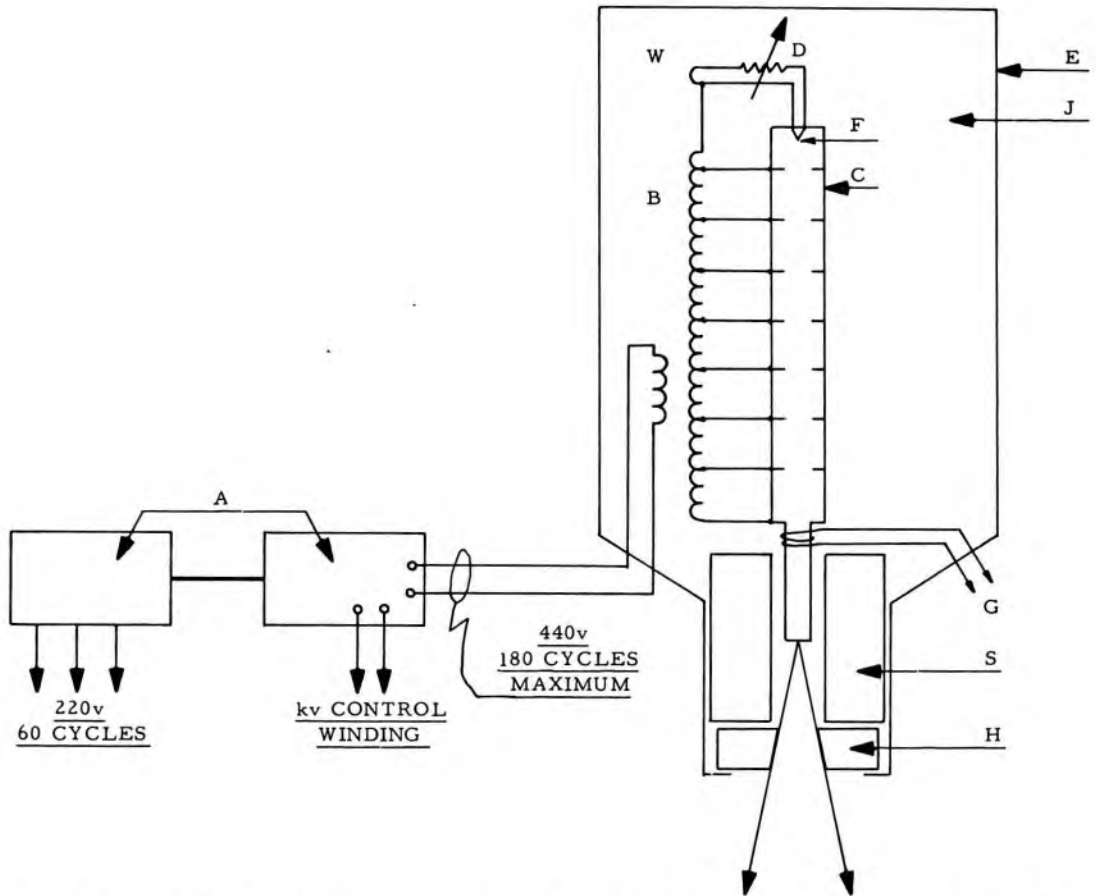


Fig. 3. Essentials of resonant transformer 1 and 2 pMV generator. Parts (see text): A. Motor generator set produces 180 cycle voltage. B. High voltage transformer produces 1 or 2 pMV between filament F and grounded target. C. X-ray tube has porcelain annular spacers and ring electrodes. Taps on transformer apply voltage in spaced steps to uniformly distribute voltage stress on porcelain insulators. D. mA reactor control for filament (remotely adjusted). E. Steel tank. F. Filament. G. Focusing coil controls focal spot size. H. Adjustable shutters control field size. J. SF_6 gas at about 4 atmospheres, for voltage insulation. S. Lead shielding.

2. *Filament.* The filament is operated at the full high voltage with respect to ground. It is heated by voltage from a supplementary winding (W) in the transformer secondary. The mA adjustment is accomplished by a remotely controlled inductive reactor (D).
3. *Gas insulation.* The tank is filled with silicon hexafluoride at about four atmospheres pressure.

A special rotary converter system (A) is provided to supply 180 Hz power to the transformer. The unit is designed so the transformer secondary is "tuned" by its associated incidental capacitance (in the windings, to the tank, etc.). This tuned, or resonant condition is easier

to achieve using 180 than 60 Hz power, for reasons beyond the scope of this discussion. The "resonant" operation minimizes the required primary power consumption. Even so the power is quite great. For example, $1 \text{ mA} \times 2 \text{ MV} = 2 \text{ kW}$ is dissipated *at the target*. Were the system not resonant, the supply power drain would be several times greater than this.

VAN DE GRAAFF UNIT.⁸ Van de Graaff generators (High Voltage Engineering Corp.) are smaller and less expensive than resonant transformer machines, making them more convenient for hospital use. In addition, constant potential operation permits operation with less tube power ($\frac{1}{4} \text{ mA} \times 2 \text{ MV} = \text{only } 500 \text{ watts}$); hence an ordinary single phase 240 V line is adequate. Van de Graaff generators are indispensable to basic physics research since they produce monoenergetic electron or ion beams with great precision of energy and current, in the range of 1 MeV to about 15 MeV. However, machines become rather large above 2 or 3 MeV since electrical insulation requires that the column length (Fig. 4) increase as the *square* of the desired voltage (i.e., a 6 MeV column is *9 times as long* as a 2 MeV column.)

Figure 4 shows the basic parts of a medical Van de Graaff machine. The x-ray tube and voltage generating system are enclosed in a heavy steel tank; electrical insulation is provided by a nitrogen-carbon dioxide gas mixture, at 20 to 25 atmospheres pressure. As in the resonant transformer system, the x-ray tube has a water-cooled, grounded anode system. The tube is made in many sections, consisting of annular shaped highly polished aluminum discs, separated by Pyrex glass insulators, all sealed together with vinyl acetate. Voltage is normally uniformly distributed along the tube by a high resistance voltage divider (VD). The x-ray tube disc hole diameters are varied and the filament system specially designed to focus the beam down to less than 3 mm diameter at the target. Tube filament current is provided by a generator mounted on the upper pulley; mA control is accomplished by means of an adjustable resistor R, which is controlled remotely by a control on the machine panel.

OPERATION. The generator provides a conveyor belt (B) to bring charges to the top; these charges are removed from the belt by a collector screen (CS), by gaseous discharge.

How are the charges supplied to the belt? The lower pulley is driven by a powerful motor, but insulated from it. A small high voltage supply raises the voltage of the insulated pulley adjustably from zero to about 50 or 60 kV constant potential. This is varied to control the charge traveling up the column. High voltage on the pulley creates a strong electric field which produces a gaseous discharge from the

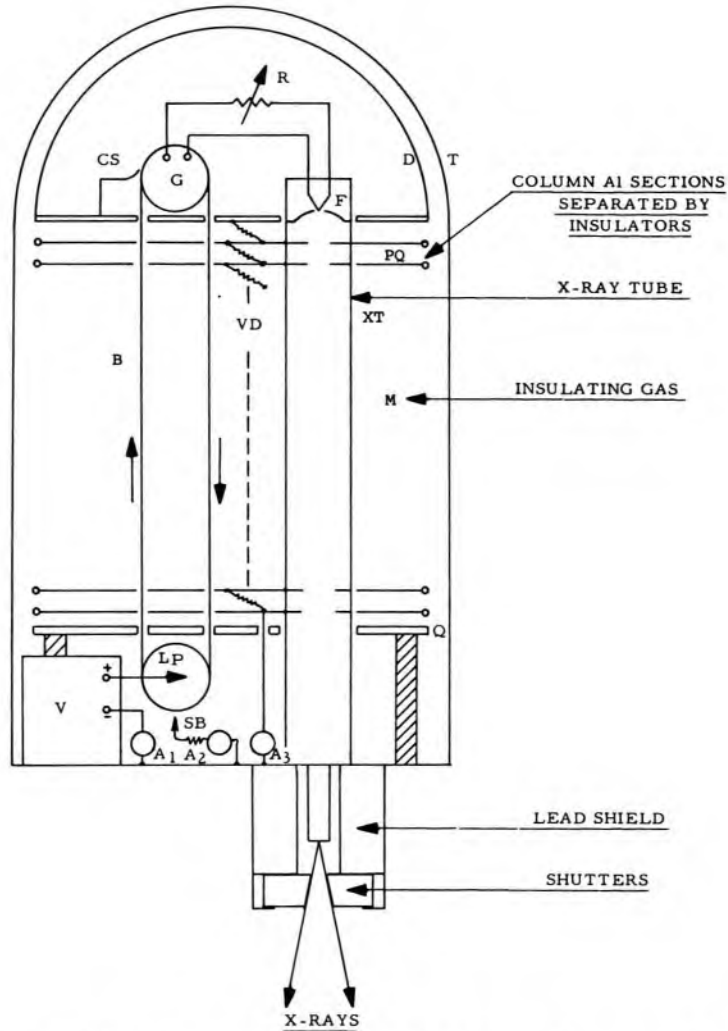


Fig. 4. Essentials of Van de Graaff 1 or 2 MV unit (see text). T. Steel tank. M. Insulating gas mixture, $N_2 + CO_2$, at 20 to 25 atmospheres. XT. X-ray tube with water-cooled, grounded anode. VD. High resistance voltage divider equalizes steady voltage across tube and column insulators. F. X-ray tube filament. G. Generator on upper pulley provides voltage to filament. R. Adjustable resistor, remote controlled, for mA control. B. Conveyor belt to transport charges to dome. CS. Collector screen removes belt charges by gaseous discharge. LP. Lower metal pulley drives belt (pulley is insulated from drive motor). SB. Spray bar. V. Adjustable high voltage supply, to produce strong electric field between spray points of spray bar SB and pulley LP. A_1 , A_2 , and A_3 are monitoring meters, actually on the control panel. PQ. Series of flat Al plates separated by insulators; plates are coplanar with tube electrodes. This is the "column."

sharp points of the grounded "spray bar" (SB) to the belt. Charges are driven by the intense electric field onto the belt where they remain until removed at the top. Note that SB supplies all the belt charge carried up the column; both the belt and pulley have excellent insulation, so extremely low "leakage" current is measured by meter A_1

(20–30 mA) while something like 300 to 400 mA is sprayed by SB.

When V departs from zero volts, charges are sprayed onto the belt and rapidly transported to the dome. There they accumulate only partially because many losses of charge tend to counteract the upward current, for example:

1. Useful x-ray tube currents (up to 250 μA)
2. Column voltage divider (about 50 μA)
3. Corona from dome to tank (10–20 μA normally).

Note all these currents increase with voltage, so the actual voltage is a compromise or steady-state value for the particular conditions. In practice, if the tube is not gassy or the column leaky the system can be remarkably stable in operation at 250 mA and 2 MV dc, particularly with special stabilizer and control circuits employed. (In practice a special electrostatic voltmeter system measures and stabilizes the terminal voltage to that desired.)

Insulation is critical in any high voltage unit such as this. Insulation of the pulley high voltage supply and belt require that the system be very dry.

The column surrounding the tube consists of aluminum plane sections separated by glass insulators. These sections are coplanar with the corresponding x-ray tube parts. Used with the voltage divider, this arrangement provides electrical field uniformity for the tube. The most critical insulator is the x-ray tube. In addition to the vacuum, the glass insulation must be very good. Any spark discharges in the column or belt can produce transient high voltages across the glass tube sections, producing sparkthrough. Any resultant exposure of new glass surfaces can potentially damage the vacuum. A special sealed "getter" type pump is included in the anode assembly of the tube to maintain the tube vacuum, thus assuring relatively long tube life.

OTHER ACCELERATOR SYSTEMS. All systems so far described use a vacuum tube across which is impressed high voltage from a separate generator. The charged particles can fall through the applied voltage only *once*. Hence, the developed MeV for a singly charged particle can never exceed the applied MV , so the attainable beam energy is limited by the tube and generator insulation capabilities. As indicated previously, the required insulator thickness rises as the *square* of the voltage. This basic physical limitation makes Van de Graaff generators enormous in size at 10 MeV and above, so a basic new approach is required in the multi-MeV energy region.

The answer to the problem lies in the use of two procedures, often combined:

1. Accelerate the charged particles in small *repeated* steps rather than in a single step.

2. Use induced or electromagnetic wave voltages to accelerate the particles, in specially designed systems.

Some machines employ *circular* motion of the beam to assure proper repetitive positioning of the particles in accelerating fields (betatron and cyclotron); in other cases, a synchronized electric field can "follow" the beam down a tube (linear accelerator). We shall now discuss the betatron and linear accelerator.

BETATRON.⁹ Figure 5(Top) shows the essential parts and operation of a betatron. The unit is basically a transformer. The primary windings are wrapped around a heavy steel yoke with poles P and P'; the dashed lines indicate the paths of the resulting magnetic flux. Note the magnetic field (arrows) is most intense in the center, falling off towards the edge where the pole shape causes a very rapid reduction in field strength. The peculiarly shaped inserts at the sides of the poles (T and T') are the betatron tube shown in cross section. This is toroidal or doughnut-shaped, and not unreasonably called a "donut" [Fig. 5(Center)]. The tube is of course well evacuated. The electrons are provided by a hot filament with its own separate accelerating circuit to raise the beam energy to about 50 keV. In addition, means are provided so that electrons enter the tube itself (are "injected") at just the proper time and for a very short interval. A transmission target is provided for x-ray production.

In operation, the primary voltage is supplied by a 180 cycle generator, resulting in a roughly sinusoidal magnetic field through the donut [Fig. 5(Bottom)]. This field steadily increases during the first quarter cycle (A through B) so that a wire around the equilibrium orbit would develop a voltage of constant polarity, with a maximum voltage at time A. An electron in this path in a wire would be slowed down by collision with atoms; in the donut, electrons are acted upon by an electric field in a manner similar to those in a wire placed there, but with *no* friction. Hence, they are continuously accelerated during the time AB. The acquired energy depends primarily on the maximum magnetic field strength, the equilibrium path area, and the time PQ during which the electron moves.

Three problems exist that we have not covered here: how the electron gets into the "equilibrium orbit," why it obligingly stays there for many *millions* of revolutions, and how we summon it to leave at the appropriate time to strike the target. These all involve considerations of magnetic systems not previously described, and the reader is referred to the references for a fuller discussion of the subject.^{9, 10} It should be noted the machine output is in the form of repetitive pulses, each of relatively constant magnitude and very short duration.

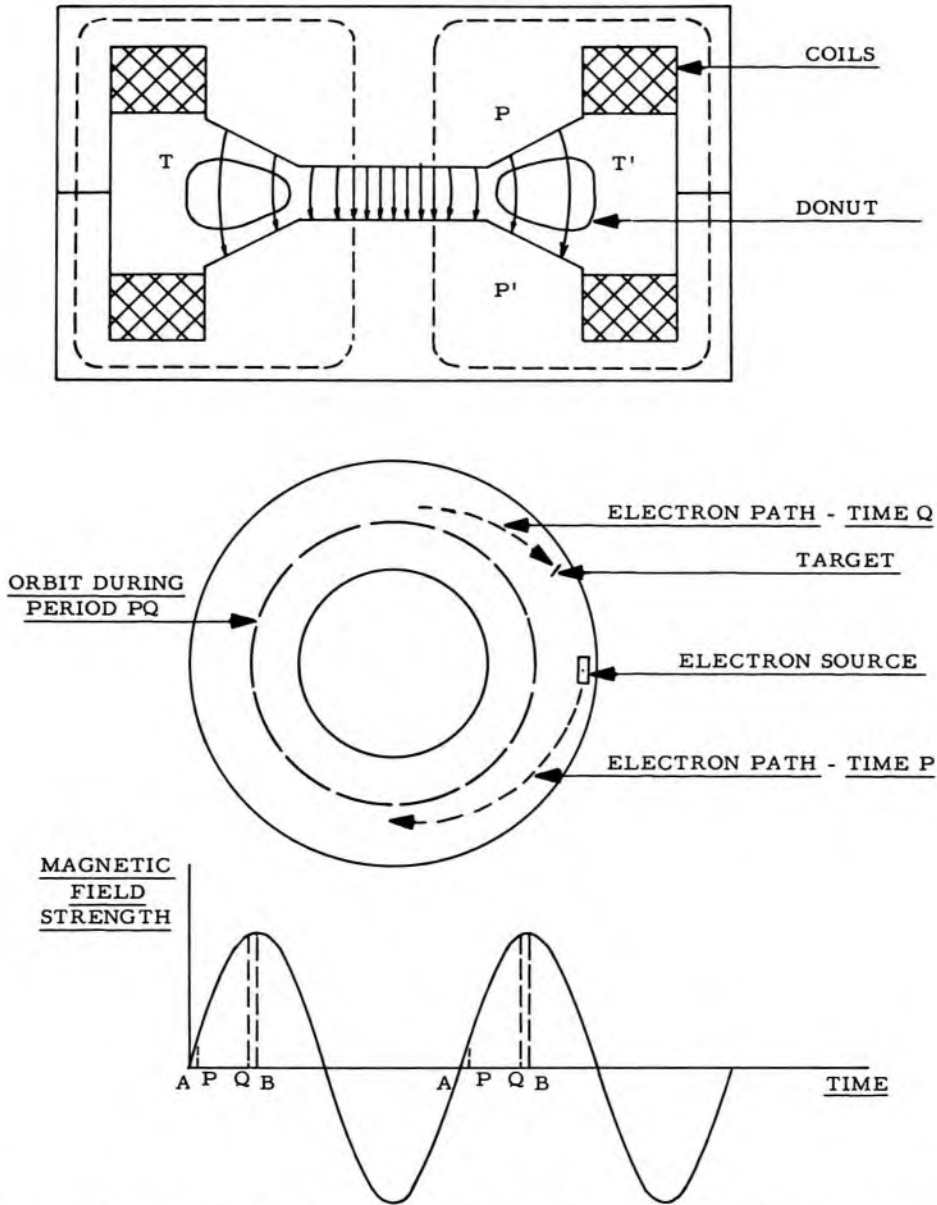


Fig. 5. Essentials of a betatron (see text). Top. Essential parts and operation. Primary windings are shown cross-hatched, wrapped around the core of a large electromagnet with poles P and P'. Dashed lines show magnetic flux path. Field falls off at edge of poles, towards donut T and T'. Center. Betatron tube ("donut"), showing electron paths (magnet and coils not shown). Bottom. Variation of magnetic field with time. P is the time of electron injection (right after A). Q is the time of electron removal (just before B).

Betatron can provide electron as well as x-ray beams for radiotherapy. In one design the same tube is used to supply both beams, with the same injection unit for each option. For electron beam operation a special deflecting magnetic field called a "peeler" is used to direct the electron beam to a beryllium window. The emerging electron beam

is of very tiny diameter and very dangerous unless the beam cross-sectional area is greatly enlarged. This is usually accomplished by the use of metal scattering foils. Suitable metal and plastic cones are provided to obtain desired field sizes.

A few comments concerning the x-ray beams are in order at this point. As previously indicated (Chap. 7), conical field flattening filters are required in all high energy x-ray generators to compensate for the rapid falloff of beam intensity away from the central ray. The required central ray attenuation for large fields is very great at 24 MV. Consequently, 24 MV betatron output rates after beam flattening may be comparatively low.

Electron synchrotrons have been employed to obtain electron beams with extremely high energies. Such machines employ synchronized electric fields in addition to betatron action to accelerate electrons in a betatron donut.¹¹ They are, however, primarily of research interest, and their clinical use has not been extensive.

LINEAR ACCELERATOR.¹² The linear accelerator has in recent years become quite popular for the generation of 4 to 8 MV x-rays. The use of intermediate photon energy makes field shaping simpler and the beam flattening filter less critical than at 24 MV. In addition, 4, 6, and 8 MV units, which permit beam rotation about the patient, have

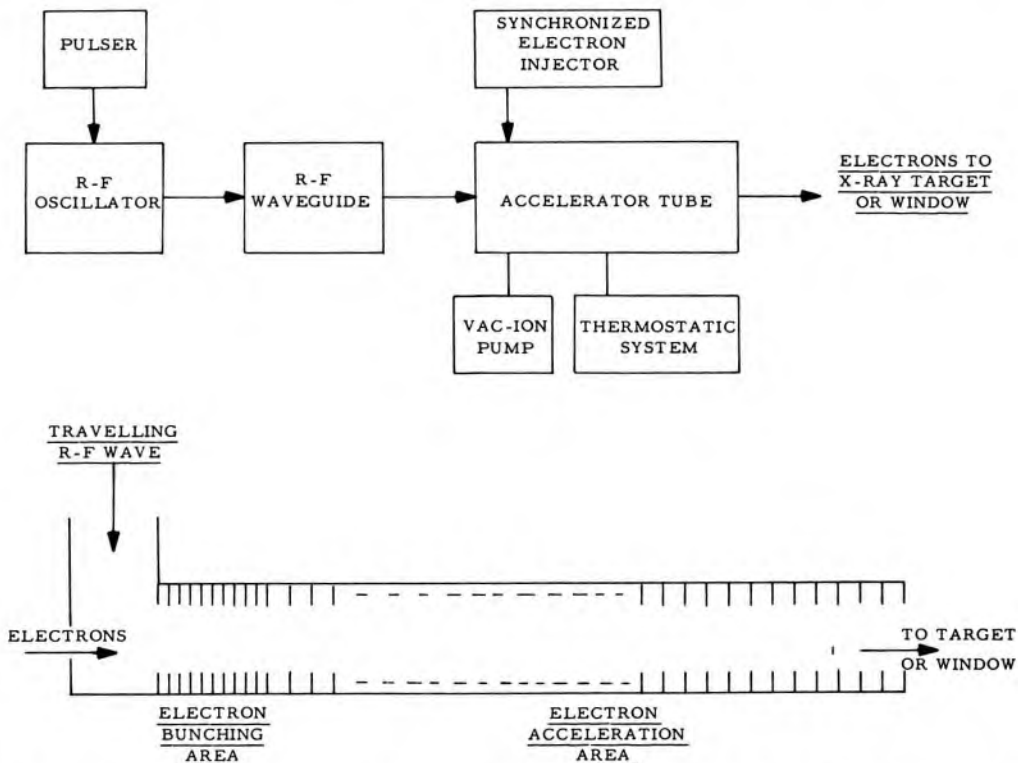


Fig. 6. Essentials of linear accelerator (see text for explanation). Top. Essential components. Bottom. The accelerator tube.

been built. Many therapists prefer rotating the machine instead of the patient because of apparently greater reproducibility of internal organ locations.

Figure 6(Top) indicates the essential system components. The accelerator tube [Fig. 6(Bottom)] receives the following:

1. A pulse of about 70 to 90 keV electrons, of very short duration, which travels down the tube in a tightly packed bunch.
2. A traveling pulsed radio wave, coupled by a special tuned wave guide structure from the radio frequency (RF) oscillator to the tube.
3. A "vacion" pump, to maintain extremely high vacuum.
4. An elaborate cooling system, required to temperature regulate the tube. If this is not provided, dimension changes from temperature variations can de-tune the system, greatly reducing the output or even rendering the unit totally inoperative.

Generally speaking, two basic actions occur in the tube. First, the groups of electrons are compressed into a small geometric volume by action of the radio wave and various magnetic field coils (not shown). Second, the electrons are then accelerated to full energy by the radio wave during their journey down the tube, in a manner similar to that of a surfboard riding an ocean wave.* The RF wave, the wave guide,

TABLE 3. INCREASE IN MASS OF THREE TYPES OF ATOMIC PARTICLES WITH INCREASE IN THEIR SPEED*

Kinetic Energy	Electrons		Protons		Deuterons	
	v/c	m/m ₀	v/c	m/m ₀	v/c	m/m ₀
0.1 MeV	.5483	1.196	.0147	1.0001	.0098	1.00005
0.2	.6954	1.391	.0208	1.0002	.0145	1.00011
0.5	.8629	1.979	.0326	1.0005	.0230	1.00027
1.0	.9411	2.957	.0465	1.0011	.0327	1.00053
2 MeV	.9791	4.914	.0657	1.0021	.0462	1.00107
5	.9957	10.78	.1026	1.0053	.0730	1.00267
10	.998817	20.57	.1451	1.0107	.1033	1.00533
20 MeV	.999689	40.14	.2033	1.0213	.1456	1.01066
50	.999949	98.85	.3141	1.0533	.2296	1.02666
100	.999987	196.70	.4283	1.1066	.319	1.05332

*[Calculated using (6-2) of Chapter 16.]

* As their speeds approach the velocity of light, electrons increase primarily in *mass* rather than *speed* as we add to their energy because they cannot travel faster than light. For example, 1 MeV electrons have 86 percent the speed of light and almost *three* times their rest mass! At 6 MeV, they have 99.6 percent the speed of light and almost *13* times their rest mass. This remarkable behavior is consistent with Einstein's restricted theory of relativity. (See Table 3 and Figure 7.)

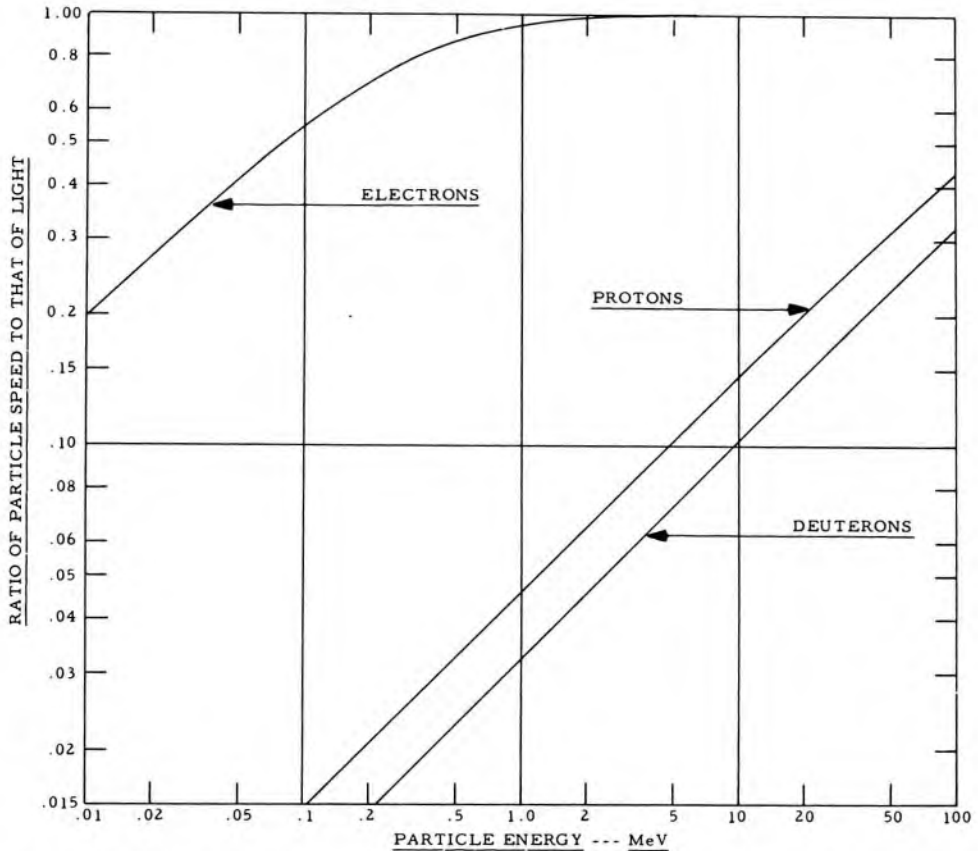


Fig. 7. Ratio of particle speed to that of light, for assumed kinetic energy. Curves have been calculated for electrons (upper curve), protons (middle curve), and deuterons (bottom curve). Note electrons reach 25 percent of light velocity near 20 keV. Protons require almost 20 MeV, deuterons, 40 MeV to acquire such speed because they are much heavier particles.

and tube must all be precisely matched to properly group and accelerate the electrons along the tube. This involves a beautiful solution of many difficult engineering problems. If the RF wave or the tube's characteristics deviate even minutely from those required, operation is adversely affected.

The newer linear accelerators have been quite dependable in clinical operation despite the required precision of adjustment.

Ion Accelerators

Cyclotrons have been used for many years to accelerate ions to energies of from 26 MeV (protons) to 200 MeV (oxygen ions). Modifications of cyclotrons have been used on a research basis for higher energies, and recent developments have already achieved fantastically great proton beam energies (33 BeV) with promise of even greater values to come.

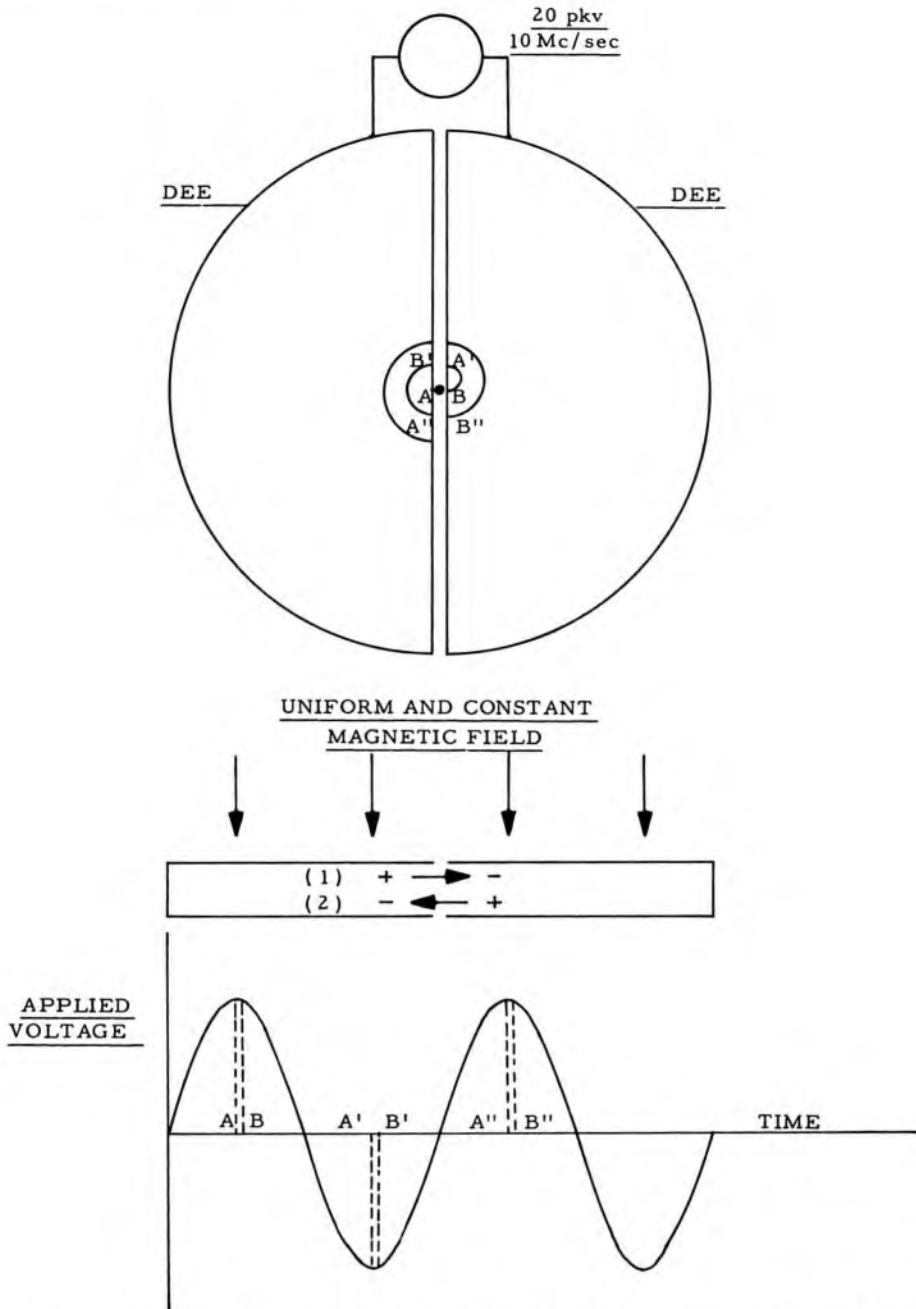


Fig. 8. Essentials of cyclotron (see text). Top. Arrangement of "dees" and applied ac voltage. Note path of particles in increasing diameter semi-circular paths. Center. Dees shown in lateral cross section. Bottom. Phasing of the ac voltage and particle emergence from the dees.

We are most interested here in the cyclotron, which has current medical applications. Consequently, high energy proton accelerators are only briefly considered.

CYCLOTRONS—DESCRIPTION (Fig. 8).¹³ The system includes the following parts:

1. A pair of hollow semicircular D-shaped metal containers (called "dees"). These are mounted in coplanar fashion and aligned as shown in an evacuated container, with a short gap separating them.
2. A very constant and uniform magnetic field directed perpendicular to the dees. This steers any traveling charged particles in semicircular paths within the enclosed volumes.
3. An appropriate ac voltage, of the order of 10 million Hz and 3 to 20 pkV, applied across the dees.
4. A source of ions, usually protons or deuterons, injected near the geometric center between the dees.
5. Target material to be bombarded by the high speed ions near the outer edge of the dees, for isotope production (target not shown).

CYCLOTRON—OPERATION. Consider the journey of a positive ion injected at A, just when the ac potential is maximum, as shown. The ion quickly reaches the right dee at B, where the electric field abruptly ceases. Since it travels with constant speed within the dee, the ion is directed by the magnetic field into a circular path and re-emerges as indicated at A'. It can be shown that time spent inside the dee by the ion is independent of its energy. The ac frequency has been selected so that during the time required to reverse the ion the voltage polarity reverses completely [Fig. 8(Center) and (Bottom)], and the ion now receives another boost in energy traveling from A' to B'. Following this it is again turned around, this time with a larger diameter path, and the process repeated.

Note this sequence can be repeated millions of times, until the ion reaches the outer edge of the dees. The target or the exit window lies here, awaiting the arrival of the bunch of ions.

CYCLOTRON—UPPER ENERGY LIMITATIONS. When particle velocities are small relative to the velocity of light, the maximum attainable ion energy depends on the dee diameter and the peak magnitude and frequency of the accelerating voltage. The dee diameter and ac frequency help determine how many reversals are permissible; the accelerating voltage, the energy acquired during each reversal.

A simple relationship exists for tuning the system:

$$f = \frac{1}{2\pi} (e/m) B \quad (16-1)$$

where f is the ac frequency used

e/m is the ratio of charge to mass of the accelerated particle

B is the constant magnetic field which steers the particles in circular paths.

Note that B must be carefully adjusted for a given type of particle (e/m) and frequency (f).

As they increase in velocity, all particles increase in mass and become more and more difficult to accelerate. In (16-1), m increases when the particle speeds approach that of light because:

$$m = \frac{m_0}{\sqrt{1 - (v/c)^2}} \quad (16-2)$$

where m_0 is the mass of the particle at rest
 m is the mass when it travels with velocity v
 c is the speed of light.

This relationship applies to *any* particle.

Table 3 presents computed values of (v/c) and (m/m_0) ratios for electrons, protons, and deuterons of various kinetic energy values. Figure 7 shows the (v/c) data plotted versus particle energy. (Use has been made in these calculations of the relationship $T = mc^2 - m_0c^2$, where T is the kinetic energy; this relationship is perfectly general and reduces to the familiar $T = \frac{1}{2} m_0v^2$ for low values of (v/c) . It is evident that electrons acquire very high velocities at relatively low energies (1/5 the speed of light at only 10 keV!). As a result, they correspondingly increase rapidly in mass, doubling at 0.51 MeV. Consequently, electrons cannot be readily accelerated in any *tuned* device like a cyclotron where mass must be reasonably constant. However, as we have seen above, use is made of the acquisition of substantially the velocity of light by electrons at a few MeV to "bunch" or group electrons in a linear accelerator.

Even protons and deuterons change enough in mass to cause desynchronization in simple cyclotrons in the range of 10 to 50 MeV. From (16-1) it is evident that two quantities can be varied to compensate for changes in m . First, the frequency can be *decreased* as m increases to maintain tuning. Cyclotrons which use this principle are called "synchrocyclotrons" since f is synchronized with m . They are used to accelerate protons to energies from 20 MeV to 730 MeV.¹⁴ The second expedient is to increase B near the edges of the dees, the location where the particles have attained relativistic speeds. A unit employing this principle is the "isochronous cyclotron"; such units have been used to accelerate protons up to at least 850 MeV.¹⁵

Newer proton accelerators now operate in the *multi-billions* of eV. They use what is essentially a large tubular system with appropriate steering magnetic fields to confine the beam to the tube axis, with synchronized ac electric fields for acceleration. Such units are used for basic physical research.

Neutron Sources

Neutron beams have both tracer and therapeutic applications in addition to radionuclide production. Tracer use depends on neutron capture by irradiated materials to produce radionuclides measurable and identifiable with scintillation and other radiation detectors. This technique ("neutron activation analysis") is possibly the most sensitive method of analysis for many elements, with advantages including speed and insensitivity to chemical form or valence.¹⁶ Therapeutic applications involve the use of both fast and very slow neutron beams. We shall comment on the three neutron sources most commonly employed for these purposes: alpha-beryllium sources; large particle accelerators, using ^2H or ^3H in the targets; and nuclear reactors.

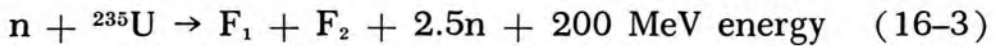
ALPHA-BERYLLIUM SOURCES.¹⁷ Alpha-beryllium units contain an alpha emitter such as radium, radon, or americium in intimate contact with ^9Be . Neutrons are produced by the $^9\text{Be} (\alpha, n) ^{12}\text{C}$ reaction. About 25,000 neutrons/sec are produced per mCi of radon or radium used. The neutron beam is readily collimated by a paraffin tube, and lead shielding removes gamma rays produced by the radium or radon. Newer high-activity alpha sources have greatly increased the neutron outputs available, in so-called "neutron howitzers."

ACCELERATOR SOURCES. High energy accelerators can be used to produce fast neutron beams by (d, n) reactions in targets on which ^2H or ^3H has been adsorbed. Units with outputs of 10^9 to 10^{11} neutrons/sec are available. [An alternative (γ, n) reaction is also possible.] Such fast neutron beams are of great potential clinical interest because their high LET likely involves minimal oxygen effect. Much research activity is in prospect to evaluate the potentialities and practical limitations of fast neutron beam therapy.¹⁸

It should be stressed that fast neutrons do not themselves produce ionization because they are neutral particles. However, they interact bodily with atoms. In such interactions they can convey considerable energy to nuclei, so the latter traverse matter with high velocity. Because they are *charged*, such nuclei can produce ionization; because they are *heavy*, this ionization is very densely distributed—i.e., the LET is great, with typical RBE values in the range of 2 to 10 times. On the average fast neutrons lose about half their energy in collisions with hydrogen atoms. Due to the relative abundance of hydrogen in soft tissue, about 90 percent of the energy of fast neutrons is absorbed this way.

Many problems must still be solved, including both dosimetry and design and collimation of sources to employ fast neutron beams for radiotherapy.

FISSION.¹⁹ Certain materials such as ^{235}U , ^{233}U , and ^{239}U undergo fission by interaction with slow neutrons. Of the three isotopes, only ^{235}U occurs naturally, as 0.7 percent of natural uranium. Its reaction is essentially as follows, on the average:



Here an atom yields two fragments of intermediate atomic mass F_1 and F_2 (atomic mass number A values from 90 to 140), between 2 and 3 neutrons, and an enormous amount of energy which appears as kinetic energy of the particles produced.

The neutrons themselves emerge with average energies of about 2 MeV. This is much too high for efficient capture by other scattered ^{235}U atoms, so fission occurs naturally at a negligible rate.

However, if one concentrates ^{235}U by separating it from the ^{234}U and ^{238}U with which it is normally diluted, it can explode spontaneously under proper circumstances. Under these conditions enough neutrons are captured to replace those leaving the system; increasing numbers of atoms therefore join the reaction, which then proceeds until the explosion disperses the ^{235}U . Synthetic materials ^{233}U and ^{239}Pu are made in reactors by neutron capture from naturally occurring ^{232}Th and ^{238}U ; these behave similarly to ^{235}U and can be used in a similar way for energy production and explosives. If they are of appropriate size and shape, two or more smaller pieces of fissionable material can be safely stored apart but then made to explode by simply bringing them together in a suitable manner. This is a principle used to detonate fission or "atom" bombs.

One speaks of a self-sustaining fission reaction as a "chain reaction." Once started in a bomb, the chain reaction is uncontrolled and cannot be stopped at will. In a nuclear reactor the process is restrained by special design. The neutron supply is carefully controlled and just enough is allowed for the task at hand; in addition, temperature is controlled by coolants.

The self-sustaining operation of either a nuclear reactor or bomb occurs at a critical juncture when the neutron supply reaches that required to sustain the reaction. This condition is often referred to as "criticality."

CONTROLLED FISSION. The basic essentials of a nuclear reactor are shown schematically in Figure 9. Fuel rods (R) are distributed regularly in a matrix (M) of a low atomic number material such as $^2\text{H}_2\text{O}$ or graphite (called the "moderator"). The fuel is usually ^{235}U or other fissionable material as metal or oxide in aluminum tubes. The neutrons escaping from the rods are slowed down rapidly by collision

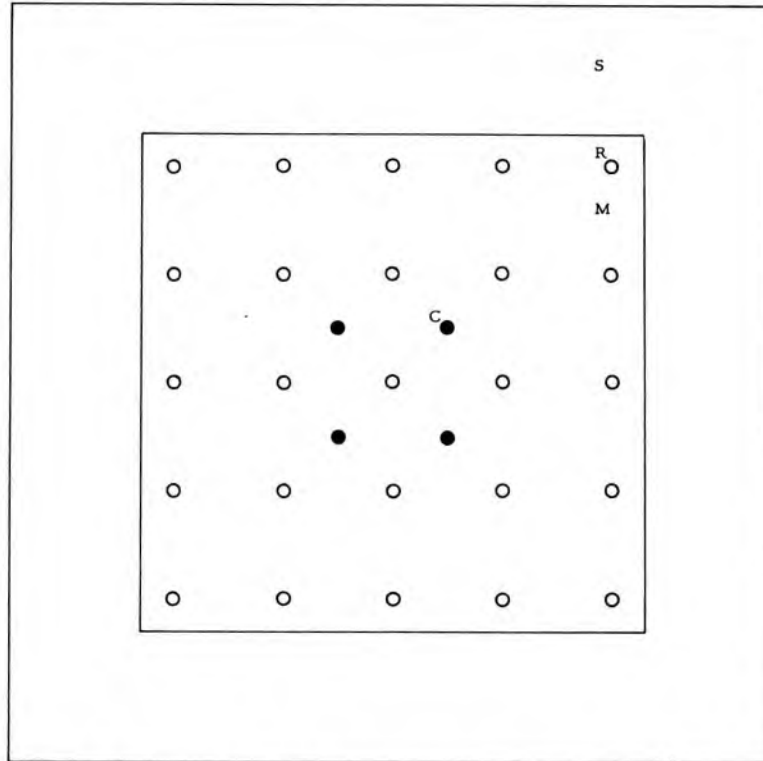


Fig. 9. Essential parts of nuclear reactor. R. Fuel rods. M. Moderator. Matrix of low atomic number material like ^{12}C (graphite) or $^2\text{H}_2\text{O}$ (heavy water) slows down fast neutrons from fuel rods to more efficient thermal speeds. C. Control rods. Can be adjustably inserted or removed, to desired depth; control rate of fission by controlling neutron flux (only four shown for simplicity). S. Shielding material. *Not shown*: Cooling means, to control reactor temperature and extract useful heat energy to make electrical power by steam-powered generators.

with ^2H or ^{12}C nuclei of the matrix. Many collisions are required to reduce the neutron energy from an average of 2 MeV initially down to 1/40 eV (114 collisions on the average with carbon); consequently, the materials must be extremely pure, lest impurities capture neutrons before they can be used. Slower-speed neutrons are readily absorbed by ^{235}U in adjacent rods with release of more neutrons, and the process builds up rapidly.

Shielding materials (S) must surround the reactor to protect both personnel and environs. Steel or concrete are useful to absorb gamma rays released by the fission products which rapidly accumulate, and concrete is a good shield for the neutrons because of its water content. Cooling is usually provided by flowing water; gas; or liquified hydrocarbons, bismuth, sodium, and certain salts. Sometimes elaborate heat exchanger systems are needed to minimize radioactive contamination of air and water.

A nuclear reactor can explode if the reaction rate becomes excessive. Control is accomplished by "control rods" (C) which are inserted in appropriate numbers and locations to suitable depth in the moderator. These are usually made of cadmium brass since cadmium has a very great cross section for slow neutrons. As a result, only the flux of neutrons needed to yield the desired reaction rate is permitted. (This is indicated by various neutron flux-indicating devices.) The rate can be adjusted from virtually zero to a safe upper level. Needless to say, automatic and dependable safety controls are an essential aspect of reactor design. Modern control systems employ "fail-safe" devices routinely, with redundant controls to assure safety, and great strides have occurred in reliable operation of nuclear reactors.

Table 4 summarizes relevant basic information about nuclear reactors. Nuclear reactor design, however, is a broad field with many highly developed specialized technologies, and this discussion is intended only to indicate basic principles common to all reactors.

SLOW NEUTRON RADIOTHERAPY.²⁰ Very slow neutrons from reactors have been used in conjunction with boron-10 for radiotherapy of brain tumors, relying on the $^{10}\text{B} (\alpha, n) ^7\text{Li}$ reaction. ^{10}B present in tissue interacts with neutrons to release energetic alpha particles, which are high LET radiation. Hence, if a high concentration of ^{10}B can be achieved in the tumor, a potentially high tumor dosage can be delivered.

Most foreign materials are excluded by normal brain structures. Brain tumors and some other pathologic conditions, however, break down the normal "blood-brain barrier." Consequently, tumors often concentrate significant amounts of the ^{10}B material (usually in borate form), so correspondingly substantial alpha-ray irradiation can be produced.

Two basic problems unfortunately arise in practice. First, other elements present in all tissues also absorb neutrons, reducing the available neutron supply at depth in the brain. Second, it is difficult to deposit useful large amounts of ^{10}B in tumors without toxic reactions.

ULTRASOUND DIAGNOSIS

Before the discovery of ionizing radiation, direct examination of accessible cavities was virtually the only method of visualizing visceral anatomy. X-ray diagnosis extended the physician's tools,

TABLE 4. NUCLEAR REACTORS—GENERAL INFORMATION

A. Needed Parts

1. Fuel consisting of fissionable materials:
 - (A) ^{235}U —natural (0.7%) or enriched (up to 90%).
 - (B) ^{239}Pu and ^{233}U produced in reactors by (n, γ) reactions with ^{238}U and ^{232}Th .
 - (C) Metal or oxide in Al tubes, appropriately spaced in moderator.
2. Moderator, of low Z materials, to reduce neutron speeds to thermal levels
 - (A) Heavy water ($^2\text{H}_2\text{O}$); expensive; used for small, versatile units.
 - (B) Graphite, generally used in larger units.
3. Control rods, to remove excess neutrons, thereby restraining reaction
 - (A) Cadmium brass, usually.
 - (B) Boron can also be used in appropriate form.
 - (C) Rods are inserted between fuel rods; depth varied to control reaction speed, or to turn off reactor entirely.
4. Cooling, by forced fluid flow around fuel
 - (A) Water under pressure or boiling.
 - (B) Molten bismuth or sodium.
 - (C) Fused salts and hydrocarbons.
 - (D) Gases like CO_2 and N_2 .
 - (E) Heat exchangers often used to minimize the radioactivity hazard.
5. Shielding
 - (A) Generally concrete.
 - (B) Graphite and water absorb neutrons specifically.
 - (C) Steel used vs. gamma rays specifically, for more compact assembly.

B. Other Information

1. Research reactors are versatile, of low power
 - (A) Used as neutron source and to make radioisotopes.
 - (B) Heavy water is useful; high neutron flux is possible.
2. Production reactors
 - (A) Used to make fissionable materials for reactors or bombs (^{239}Pu and ^{233}U).
 - (B) Graphite moderator.
3. Power reactors, designed for efficient electricity production.
4. Fuel rods accumulate fission products, poisoning fission reaction. Must be reprocessed often by chemical separation, involving potential hazards from accumulated fission product radioactivity, ingestion of ^{239}Pu , etc., and exceeding criticality in handling solutions of fissionable material.

and this was later complemented by radionuclide tracer scans. As previously shown, basic limitations exist in simpler x-ray examinations. They arise from the superposition of images of structures traversed by the rays and the limited available contrast levels in soft tissues. Multiple views and the various special studies have greatly increased the harvest of useful information, but many situations arise in which x-ray studies contribute in only a limited way to the solution of a diagnostic problem. Radionuclide scans are also of great usefulness but so far yield relatively coarse anatomic structural detail. The two newer diagnostic modalities have therefore been actively investigated in recent years.

Ultrasound operates on a reflection rather than transmission or emission principle. It is reflected from acoustic inhomogeneities in the body. These are caused mainly by differences in elasticity (stiffness), but also of density of various tissue constituents. Available instruments can be used to make detailed records of cross-sectional anatomy similar to those derived from cadaver studies. These are quickly obtained without patient discomfort or hazard, with great potential resolution of detail. Ultrasound is also used to measure blood flow by the Doppler principle.

In this section we briefly discuss some of the physical principles and the applications of ultrasound in medical diagnosis.*

Sound

NATURE AND PROPERTIES.²¹ It must be stressed that sound is a matter, not electromagnetic, wave. Its properties are therefore entirely different from those of x-rays. Sound waves are vibrations in solids, liquids, and gases. Those in the frequency range from 20 to 20,000 Hz are audible; those below 20 Hz are called subsonic and those above 20,000, ultrasonic. Diagnostic anatomic studies are usually performed using frequencies of 1 to 15 MHz while Doppler blood flow measurements are done at about 5 MHz. At these frequencies sound waves are totally reflected by gases. Sound must be conveyed to the body without air path; in addition, body air and gas cavities (such as lung) cannot be explored with ultrasound.

The speed of sound was shown by Newton to depend on the density ρ and elasticity M of the medium, as follows:

$$v = \sqrt{\frac{M}{\rho}} \quad (16-4)$$

This formula applies to homogeneous materials such as metals, liquids, and gases, but it is hard to define what the "elasticity" means

* Ultrasound has also been used therapeutically.

in a complex matrix of connective, vascular, glandular, nerve, and other tissues of the body. Fortunately the velocity of ultrasound is relatively uniform in living soft tissues, about 1,540 m/sec. This is fairly close to the velocity in pure water and normal saline solution at the same temperature. Bone is denser and much more rigid, yielding greater velocity; for example, the speed is twice as great in ivory as in water.

INTERACTION WITH MATTER. Diagnostic information is obtained from sound reflections or echoes at tissue boundaries. Two types of factors affect how large an echo is produced by sound normally incident at a boundary: the density and sound velocity in the two media. Figure 10 indicates the general relationships, with an illustrative calculation. Note at the top of the figure that soft tissue echoes are

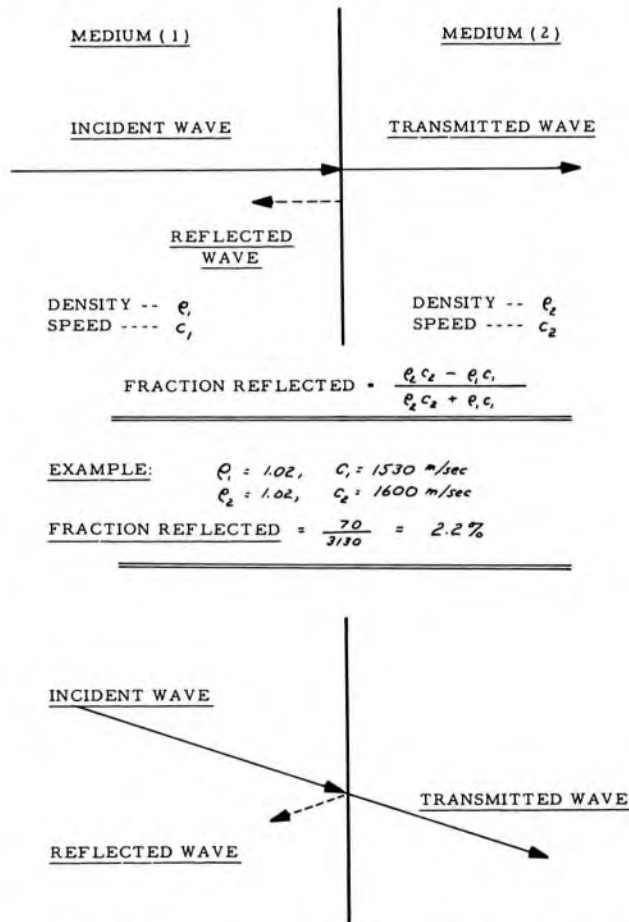


Fig. 10. Reflection of sound waves at boundaries. Top. Normal incidence. Reflected beam is returned to transducer for detection and depth indication. Formula is shown for fraction of incident beam reflected, with sample calculation. Of course returned beam is partially absorbed along the way by intervening tissue. Bottom. Oblique incidence. Obliquely incident beam is reflected away from original direction and cannot be used by the same transducer depth gauging.

often rather faint, requiring sensitive detector systems. Also, at the bottom oblique reflection leads to loss of the signal, and irregular surfaces produce weaker echoes and on occasion undesired reverberation signals. The obtaining of meaningful results requires a well-engineered system used with considerable skill and anatomic knowledge.

Sound waves are progressively attenuated as they traverse objects. The absorption coefficient varies in a rather complicated way with frequency, but a roughly proportional relationship exists in most materials. (There is an acoustic engineering rule of thumb: attenuation is about one db/cm/MHz.) This fact is important practically since higher frequency ultrasound cannot penetrate to reach deeper structures.

High intensity sound is known to be harmful at even audible and subsonic frequencies. Ultrasound at high intensities (many watts per cm^2) can destroy cells; precision focused beams have been applied to selectively destroy minute areas of the hypothalamus in treatment of Parkinson's disease. The action is primarily thermal, but at very high intensities ionization phenomena also occur.

Fortunately, very low intensity beams can be usefully employed in diagnosis. Intensities of the order of 0.005 watt/cm^2 are used for the usual echo work; in Doppler blood flow measurements, 0.05 watt/cm^2 . Experimental work offers convincing evidence that these levels are orders of magnitude below those required to produce detectable biologic effects.

Basic Diagnostic Methods

Both reflected and transmitted pulsed ultrasound beams are used to obtain anatomic and physiologic data. In addition, the Doppler principle is being studied intensively as a method of blood flow evaluation. Basic principles of these methods will be briefly discussed.

REFLECTION. Figure 11 shows the basic approach. The instrument produces an adjustable ac voltage of the desired ultrasound frequency; this is used to drive the transducer. The energizing signal is a pulse of only short duration, ranging from 1.5 to 4.5 microseconds; this is only a few cycles of one MHz current [Fig. 11(Bottom left)]. Also provided is circuitry to display all reflected echo signals on an oscilloscope, where they are commonly photographed by a Polaroid camera.

The transducer contains a crystal that vibrates in response to the applied electrical pulse. (This uses the reverse piezoelectric effect.) This vibration generates a corresponding ultrasound pulse which is communicated to the skin through water or a thick jelly to avoid excessive losses which occur at air-tissue interfaces.

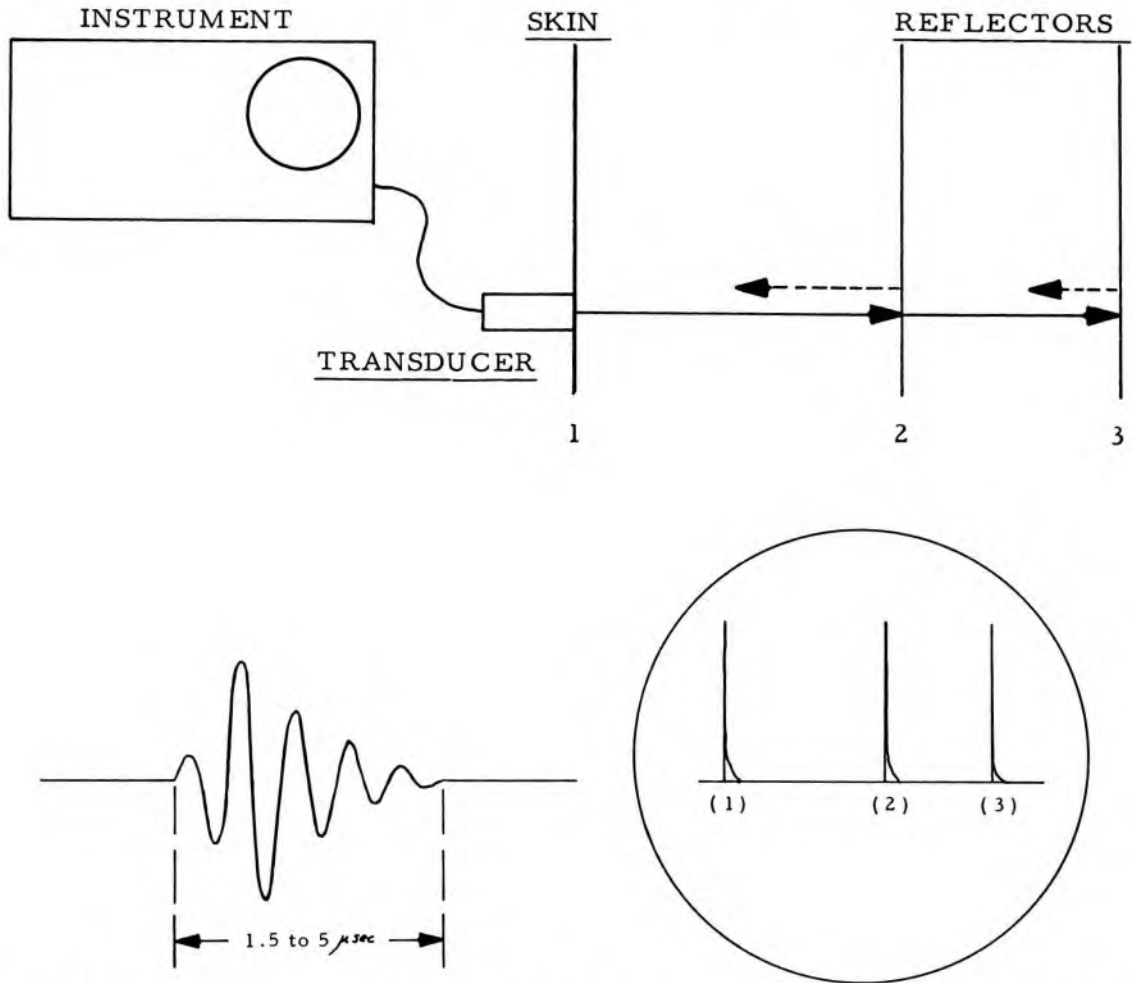


Fig. 11. Reflection type ultrasound measurement system (see text). Top. Basic operation. Bottom Left. Ultrasound pulse wave form. Bottom Right. A-mode display.

When the incident sound pulse strikes acoustic inhomogeneities, echoes are returned to the transducer. This unit, remarkably, works equally well in producing an ac *voltage* from the sound echo as *sound* from the original voltage. By the time the echo sound pulses arrive, the crystal has usually long since stopped vibrating because the initiating voltage pulse lasted only a few microseconds. For example, consider a reflection from 5 cm below the skin. The sound travels 5 cm *in* and 5 cm *back*, or 10 cm total. This takes $(0.1 \text{ m}/1500 \text{ m/sec})$ or $66 \mu\text{sec}$, much longer than the time required by the crystal to prepare to respond to the echo signal.

Three pulse signals are shown on the oscilloscope screen of Figure 11(Bottom right). The first is from reflection at the skin, the others from the two interfaces of Figure 11(Top). These are displaced in time by amounts equal to twice the surface depths divided by the average sound speed. By use of proper circuits, the pulses may be displayed as shown, the echo numbers corresponding to the reflecting

surfaces that produced them. If bone or other material of atypical sound velocity is traversed, some correction is required to validate the indicated positions. This correction is made easy to perform in newer units.

Transducer design is as yet something of an art, involving many difficult problems of acoustic focusing, coupling, and damping design. Quartz is sometimes used because of its durability, but various ceramic materials such as barium titanate yield more consistent pulse shapes.

DOPPLER FLOWMETER.^{22, 23} Whenever a source of sound travels towards or away from a receiver there is an apparent increase or decrease of pitch. When a train approaches while sounding a warning whistle, the observed pitch is *higher* as the train approaches, abruptly *lower* after it passes. The ultrasound Doppler flowmeter operates on a modification of this principle.

Figure 12 shows the instrument schematically. A unit with separately mounted transmitter and receiver transducers is placed against the patient, over the vessel whose flow it is desired to measure. Unlike the previous application, the signal is now a continuous vibration rather than short pulse. The sound is reflected by particles in the blood (such as cells) to the receiver. Because these particles are moving relative to the source, the sound frequency received and hence reflected by them is modified. The frequency shift is given by the formula:

$$f = \frac{2v}{c} f_0 (\cos \theta) \quad (16-5)$$

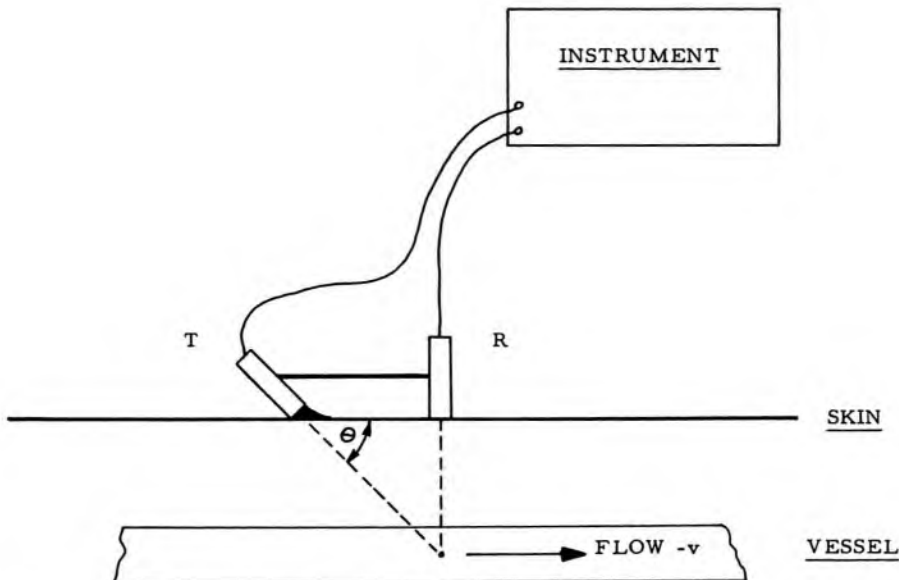


Fig. 12. Doppler flowmeter setup (see text). T. Transmitter ultrasound transducer. R. Receiver transducer. θ . Inclination to skin surface of entering beam. T and R are mechanically linked, so angle θ is fixed.

where v is the desired flow velocity in the blood vessel, c is the speed of sound, and f_0 is the original sound frequency (usually about 5 MHz). The received and original electric signals are combined electronically to obtain the desired magnitude of the Doppler frequency shift. This can be displayed graphically as a function of time. More often it is simply used to produce an audible tone, whose pitch indicates the flow. Pulsatile flow produces recognizable characteristic sound patterns. These may ultimately be employed in a manner analogous to auscultation sounds.

There are at present certain limitations to the Doppler flowmeter. First, it cannot tell one anything about the *direction* of flow, only its *magnitude*. (This is true of current designs. A frequency discriminator rather than beat frequency system could correct this deficiency.) Hence, one cannot unequivocally detect reversals of flow; they are simply indicated by a drop to zero frequency with return to higher values. Secondly, it cannot evaluate volume of flow—only linear velocity. Used intelligently, however, it can give much useful diagnostic information. A major advantage is its relatively low price (less than \$1,000).

Technical Aspects

The usefulness of an ultrasound study depends on many variables. We shall briefly discuss two basic aspects: detail resolution and information display.

RESOLUTION. Diffraction usually sets a limit on details which can be separated simply for *any* wave.* (With light microscopes the practical limit is about 0.1 to 0.2 microns, depending on the color of the light used.) The wavelength of ultrasound is:

$$\lambda = c/f = \frac{1,500,00 \text{ mm/sec}}{f(\text{Hz})} = \frac{1.5 \text{ mm}}{f(\text{MHz})} \quad (16-6)$$

For frequencies of 1.5 to 15 Hz, ultrasound wavelengths are hence about 1.0 to 0.1 mm.

In actual practice, transducers, electronics, and recorders introduce their own problems, and 0.1 to 1.0 mm resolution is at present a limit. Recall that patient and part motion as well as part roundness also introduce blur in radiography. Part motion and indistinctness of borders may also be expected to reduce the realized resolution of in-vivo ultrasound measurements.

* Some newer statistical methods can apparently overcome this limitation, using special electronic signal processing and display systems.

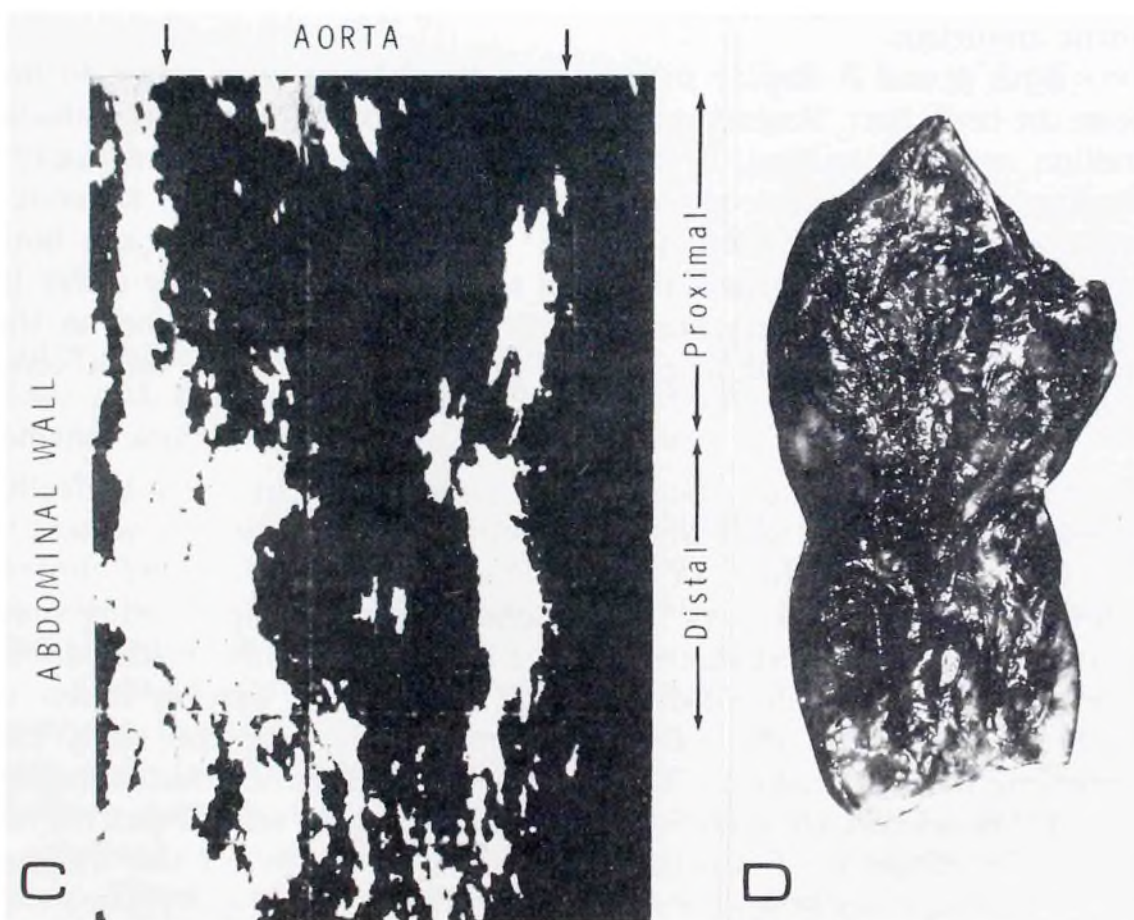
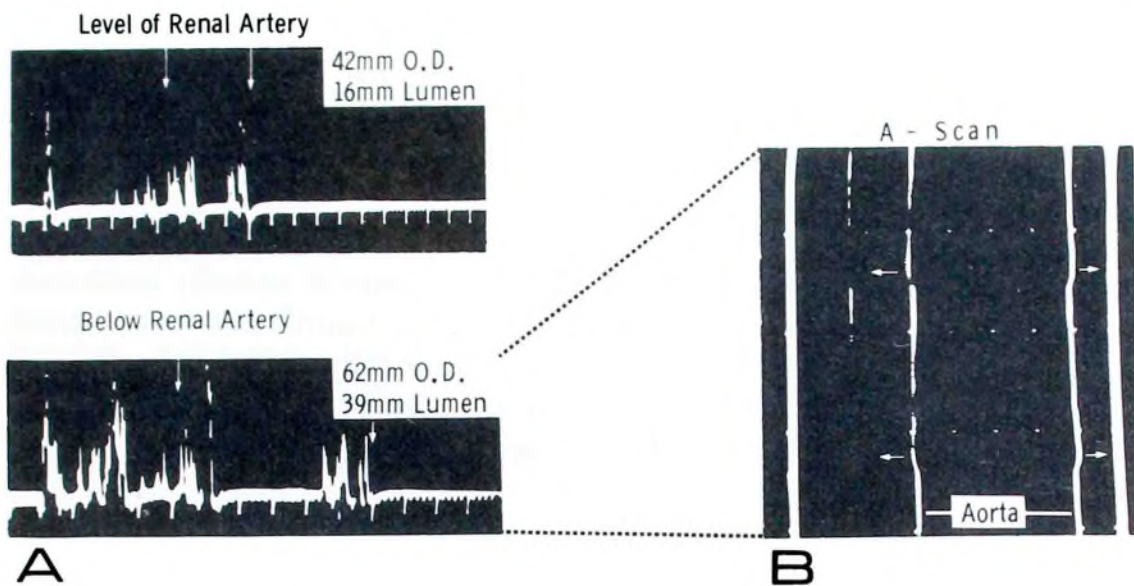


Fig. 13. Ultrasound records of aneurism of aorta. A. Two A-mode displays taken along aorta. B. B-mode display. C. B-scan of aorta—front-to-back through aorta section. D. Actual postmortem aorta specimen. (From Goldberg, B.B., Ostrum, B.J., and Isard, H.J. Ultrasonic aortography. *J.A.M.A.*, 198:353, 1966.)

DISPLAY SYSTEMS. Various terms have been developed to describe different ultrasound display systems. They are derived from radar system terminology. There are two display "modes," "A" and "B," and two main categories of scans, B and C. These will now be described in general terms.

An A-mode display is simply a record of the echo pulse sequence plotted versus time [Fig. 13(A)]. The time scale is usually calibrated in millimeters, so the positions of reflecting boundaries are readily determined. When bone transmission corrections are needed, as in echoencephalography studies, special circuits may be used to make such corrections automatically for assumed or measured thickness and density of bone.

The B-mode display records the variation of the location of the pulse peaks with time as a graph on the oscilloscope [Fig. 13(B)]. It is usually employed to indicate pulsatile motion (such as that of a large blood vessel or heart chamber). Figure 13 demonstrates an aortic aneurism.

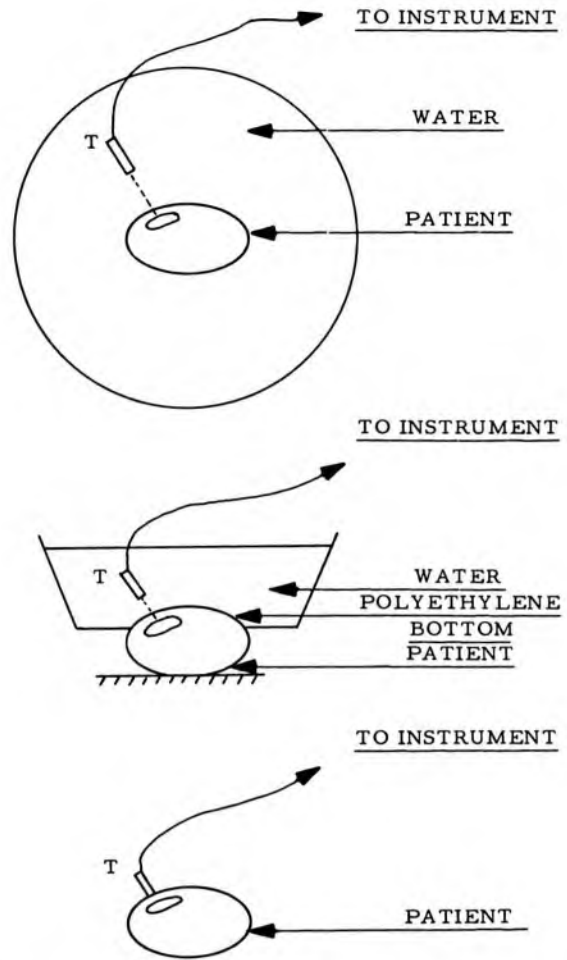
Both A and B display modes use a fixed transducer; they do *not* scan the body part. Scanning always involves some form of transducer motion over the patient, to obtain echo patterns. These are usually displayed on an oscilloscope and photographed to obtain an anatomic cross section of the patient [Fig. 13(C)]. B-scans and C-scans both give a display of the anatomic cross section. However, they differ in the method used to correlate the oscilloscope trace position to the transducer location and angle.²⁴ At present B-scans are most commonly performed.

TYPES OF B-SCANNING. Scans may be carried out with both the transducer and body part under examination immersed in water, to avoid air reflection [Fig. 14(Top)]. This is obviously a great clinical limitation. Techniques have consequently been developed using plastic bags filled with water instead [Fig. 14(Center)]. This avoids patient immersion and provides fair acoustic coupling if care is taken to avoid air bubbles in the water and between the bag and skin. The water-bag method, however, has at least three inherent disadvantages:

1. Good contact requires significant pressure on the patient. In addition to producing discomfort, pressure on the inferior vena cava frequently causes syncope.
2. Substantial acoustical attenuation occurs in the water, reducing the possible depth of penetration.
3. Considerable losses arise from attenuation of sound at the skin and water interfaces with the plastic bag.

An alternative method is contact scanning [Fig. 14(C)]. A water-soluble grease is used to assure good sound coupling, and the trans-

Fig. 14. Means of coupling ultrasound from transducer to patient. Top. Immersion technique: both transducer and patient immersed in tank. Center. Water-bag technique: transducer in water; plastic bag in good contact with patient skin. Bottom. Contact technique: transducer contacts skin through water-soluble grease, to avoid air reflection.



ducer moved about the patient in contact with his skin. This can be done with the patient in virtually any desired position, assuring more normal anatomic locations of viscera than with the water-bag technique.

As the transducer moves over or around the patient, it is usually also simultaneously rocked back and forth (arc-sector scanning). This is done to sample as many of the reflecting surfaces as possible during the sweep of the area.

Contact scanning has several advantages over the other coupling methods. First, it is very fast. In addition to reducing patient discomfort, this fact permits sampling the anatomy quickly, minimizing motion blur. Second, acoustic losses through coupling material are minimized, so useful scans can be obtained from deeper body structures. Third, contact scanning does not displace body organs significantly since parts are neither immersed nor placed beneath a heavy bag.

Applications

A brief list of some current diagnostic applications of ultrasound is given in Table 5.²⁵ Basic as well as clinical research is in progress,

TABLE 5. SOME DIAGNOSTIC USES OF ULTRASOUND²⁵

A. Echo recordings

1. Eye: tumors, foreign body localization—15 MHz
2. Circulation
 - (A) Heart valvular disease
 - (B) Abdominal aorta, follow-up of aneurism
 - (C) Pericardial effusion
3. Intracranial disease
 - (A) Midline shift (mass lesions). About 12 to 13% false positives reported
 - (B) Diencephalic width
4. Breast: mass lesions
5. Liver and biliary system: mass lesions, fibrotic disease, fatty metamorphosis
6. Obstetrics—pregnancy: single vs. multiple (early as 10 wks.), biparietal encephalometry, pelvimetry
7. Gynecology: mass lesions (solid tumor vs. cysts)

B. Doppler

1. Fetal heartbeat, placental localization
2. Circulation in vivo.

and many future changes are to be expected. The use of scans instead of A-mode display appears likely in much future work as contact scanning problems are solved.

As in the history of x-ray diagnosis, it is to be expected that new studies will be developed and old studies improved as new techniques and equipment are developed.

THERMOGRAPHY

The general body temperature is a very useful diagnostic indicator, and the clinical thermometer is a classic medical tool. Skin temperature is also a useful diagnostic indicator, but its measurement is less conveniently carried out. In the past, small thermocouples have been employed, but their use has been restricted to research because of the difficulties involved.

Thermography provides a more elegant and convenient method for measuring temperature distribution over extensive areas of the body. The procedure is carried out without contacting the body at all. Unlike x-ray, radionuclide, and ultrasonic diagnosis, no physical agent is applied to the body; in fact, it is the body's own radiant energy which

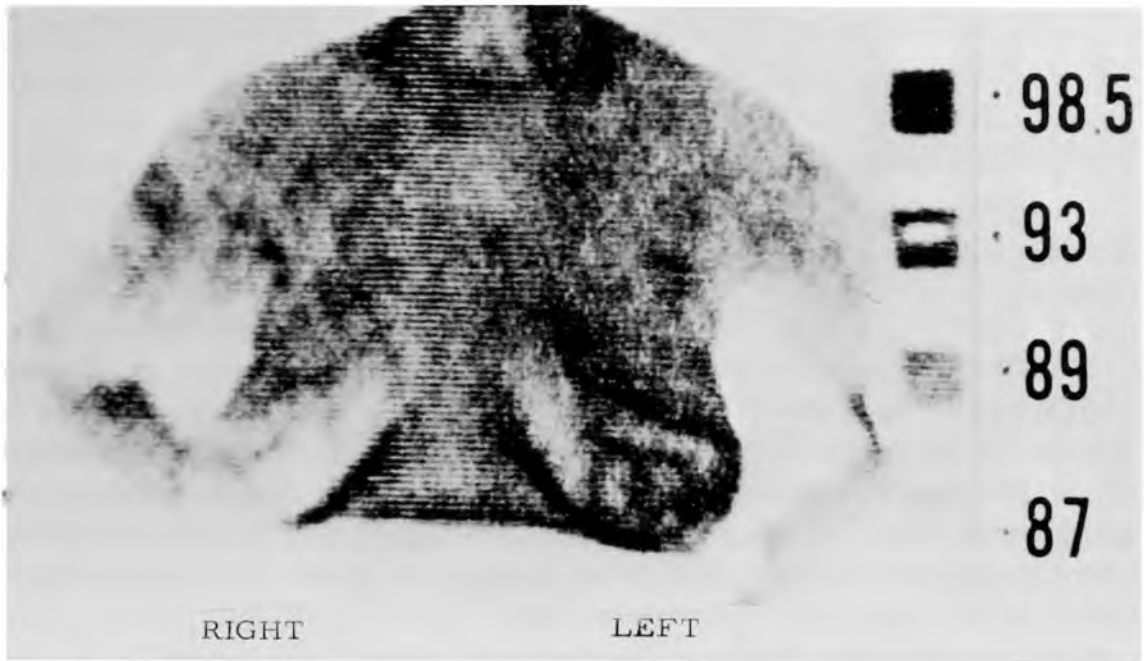
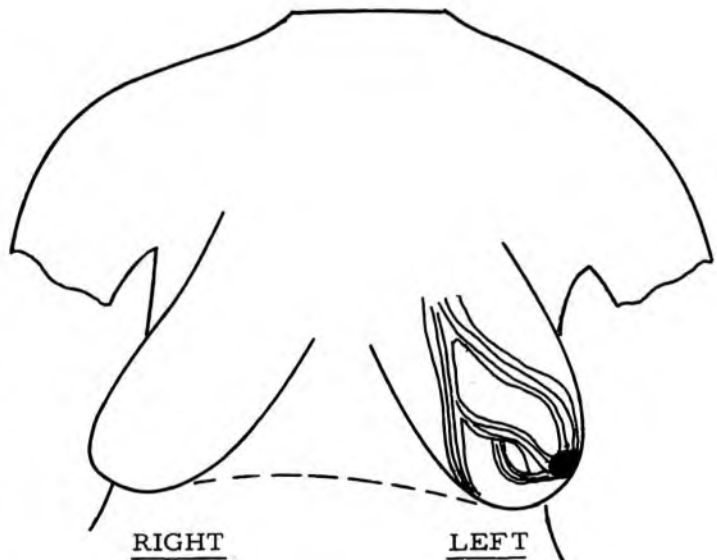


Fig. 15. Thermogram of female patient.²⁶ Patient was later shown to have 2 cm diameter carcinoma of the left breast. Note left to right asymmetry of temperature distribution. Above. Thermogram record itself. Facsimile recorder unit. Note elevated skin temperature readings on the involved (left) side. (Wallace, J. D. Personal communication.) Right. Sketch showing higher temperature areas. Not only lesion itself but also drainage areas are warm.



produces the result. The record is essentially an infrared recording of the patient. Warm areas generally appear light, cool areas dark, with gradations of gray for intermediate temperatures.* There are normal patterns characteristic of different body areas; it is the departures from these which hopefully provide clues as to subclinic pathology. To illustrate, Figure 15 shows a "thermogram" of a female patient with a later proven breast carcinoma.²⁶

* This is true in the Barnes commercial unit. The Smith Industries, Ltd., unit used by Wallace differs technically and shows warmer areas *dark* rather than light. Figure 15 was obtained using a Smith Industries unit.

We shall now briefly discuss the basic principles of thermography and indicate some reported applications.

Basic Principles

Warm objects tend to lose heat to nearby cooler objects. Three mechanisms; conduction, convection, and radiation, are commonly involved.

CONDUCTION AND CONVECTION. Conduction is illustrated by a poker with one end in a fire. The other end also becomes warm but not so hot as the heated end. Heat is transferred along the bar as a result of communication of kinetic energy from atom to atom in the bar. In general, all materials conduct heat to some extent. In winter our homes lose considerable heat through the walls, which, for this reason, often feel cold in more poorly insulated buildings. Convection differs from conduction in that fluids carry the heat away from warmer objects by their gross *motion*. The winds and ocean currents are examples of convection transfer of heat on a grand scale; it also occurs in a heated pot of water or a room with a warm steam radiator at one side.

RADIATION. Both conduction and convection require matter to carry heat from a hot object to cool ones. However, all warm objects also emit electromagnetic waves.²⁷ Until very high temperatures are reached, these waves are restricted to the very long wavelength region beyond visible red light and called "infrared waves"; at higher temperatures they become more intense and are also accompanied by visible light.

The infrared emission is related to the temperature of the object:²⁸

$$I = kT^4 \quad (16-7)$$

Here, I is the rate of emission of infrared wave energy, and T the object's absolute temperature (273.16 plus its temperature in degrees centigrade). The quantity k depends primarily on the reflective properties of the object and is maximum for a so-called "black body." The essentially matte nature of skin makes human beings of all pigmentations almost perfect "black bodies" as infrared radiation sources. The net signal received by a radiation detector above a patient depends on the *difference* in temperature between the patient and detector since the detector itself also radiates energy according to (16-7). For example, in a room at uniform temperature a detector would receive no signal from a cadaver also at room temperature. Substituting a live person, however, would result in an immediate

signal because a live person is then warmer than the detector. If an appropriate system scans the patient's skin, detector signals will be produced corresponding in magnitude to the skin temperatures of scanned locations. The detector signal is amplified and synchronously displayed on a cathode ray oscilloscope: the higher the body area temperature, the brighter the spot. A Polaroid camera is then used to provide a permanent record. (Alternatively, the Smith Industries, Ltd. system uses a facsimile recorder, but the basic principle is the same.)

PRACTICAL ASPECTS. One basic problem in any thermography unit is the very low intensity of infrared signals available. As a result, noise becomes troublesome. Practically, this results in a great increase in required scanning time (to permit averaging out noise fluctuations— analogous to counting weak radioactivity signals longer to improve statistical accuracy). Another technique is to reduce noise by cooling the detector to liquid nitrogen temperature; this procedure and the use of a solid-state detector greatly increase useful scanning speed, permitting a scan with 40,000 individual measurements in 30 seconds.

Some procedures are essential to assure accuracy and reproducibility. They include the following, as described by Gershon-Cohen:²⁹

1. A ten minute precooling of the uncovered skin at 70 to 76°F.
2. Thermostated room, with low temperature gradients.
3. For thorax scanning, arms extended, to avoid cross radiation from axillae.
4. Landmarks are useful. Small aluminum tabs on the skin have been used successfully for this purpose.

Applications

It must be stressed that results of this procedure have been reported only during the past few years, and clinically the method is not yet fully evaluated. There have been some exciting preliminary results, however. The following is a partial summary:

1. *Breast malignancy.*
 - a. *Gershon-Cohen.*³⁰ 750 patients thermographed. Finds thermography may materially advance diagnosis of breast disease.
 - b. *Swearingen.*³¹ 100 patients. Found "improved results" using thermography and mammography combined over mammography alone.

- c. *Wallace, Dodd, et al.*³² Found false positives from localized inflammation as well as functioning breast tissue surrounded by cysts.
2. *Placental localization.* Young³³ reported technical details of preliminary use.
3. *Cardiovascular.* Wood³⁴ reported a temperature reduction of ocular and forehead areas following carotid artery occlusion.
4. *Other uses reported under investigation:* rheumatoid arthritis and peripheral circulatory disorders.

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N.B.S. Handbook 85:	Physical Aspects of Irradiation (ICRU Report 10b)	0.70
N.B.S. Handbook 86:	Radioactivity (ICRU Report 10c)	0.40
N.B.S. Handbook 87:	Clinical Dosimetry (ICRU Report 10d)	0.40
N.B.S. Handbook 88:	Radiobiological Dosimetry (ICRU Report 10e)	0.25
N.B.S. Handbook 89:	Methods of Evaluating Radiological Equipment and Materials (ICRU Report 10f)	0.35

D. N.B.S. Handbooks on Radiation Protection*†

N.B.S. Handbook 55:	Protection Against Betatron-Synchrotron Radiation up to 100 MeV	0.25
N.B.S. Handbook 59:	Permissible Doses from External Sources of Ionizing Radiation	0.35
N.B.S. Handbook 63:	Protection Against Neutron Radiation up to 30 MeV	0.40
N.B.S. Handbook 65:	Safe Handling of Bodies Containing Radioactive Isotopes	0.15
N.B.S. Handbook 69:	Maximum Permissible Body Burdens and Maximum Permissible Concentrations of Radionuclides in Air and in Water for Occupational Exposure	0.35
N.B.S. Handbook 73:	Protection Against Radiations from Sealed Gamma Sources	0.30

* Available from Supt. of Documents, U.S. Government Printing Office, Washington, D.C. 20402.

† The NCRP has recently become a chartered agency which now publishes its reports independently. These include:

- NCRP Report No. 33: Medical X-ray and Gamma Ray Protection for Energies up to 10 MeV
- NCRP Report No. 34: Shielding Aspects of Medical and Gamma Ray Installations up to 10 MeV
- NCRP Report No. 35: Dental X-ray Equipment and Installations

For further information contact the National Council on Radiation Protection: NCRP Publications, P.O. Box 4867, Washington, D.C. 20008.

N.B.S. Handbook 76:	Medical X-ray Protection up to 3 Million Volts	\$0.25
N.B.S. Handbook 80:	A Manual of Radioactivity Procedures	0.50

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APPENDIX B

PITFALLS IN FILM DOSIMETRY

1. **ENERGY DEPENDENCE.** X-ray films all respond differently from air or tissue because their emulsions contain silver and bromine atoms of atomic numbers 47 and 35 respectively, as compared with about 7.5 for tissue. Hence, films differ greatly from tissue in photoelectric attenuation coefficients and respond disproportionately to low energy photons. They can yield serious errors from this cause if care is not taken in planning and interpreting measurements.

2. **LINEARITY.** Response of films is roughly proportional to dosage when they are exposed directly to ionizing radiation, *without intensifying screens*. However, the relationship is only approximately so. The response with screens is both less linear and dependent on screen response. This varies over the screen, as well as with temperature and photon energy.

3. **LATENT IMAGE FADING.** The latent image is not perfectly stable after exposure. It varies greatly during the first few minutes, then becomes more constant, fading gradually thereafter. The rate of fading is relatively great for smaller film grain, becoming extreme with very fine grain emulsions. Fading is reasonable with x-ray films: a type M-industrial film fades less than 10 percent in two weeks.

4. **FILM PROCESSING.** This is undoubtedly the greatest source of problems in film dosimetry. Extreme care is required, and it is recommended that all manufacturer's chemistry and procedure recommendations be followed strictly.

5. **FILM SENSITIVITY VARIATIONS.** Considering the complexities of film manufacture, it is remarkable how reproducible films are:

Batch to batch	± 10 percent
Film to film (same batch)	± 5 percent
Over a single film	± 1 percent

However, once the box is opened greater batch-to-batch variations may result from moisture and vapor effects.*

* Corney, G. Personal communication.

APPENDIX C

GLOSSARY

Terms have been included which are of most general interest in medical radiology. When readily available, definitions are quoted or paraphrased from references given in the text; otherwise they are the author's own.

In addition to serving the usual purpose, this glossary is intended as a study aide. Consequently definitions of related terms are often grouped together under logical headings to facilitate learning and memory. When this is done, brief definitions are also provided in the alphabetical listing, with cross-references.

ABSORBED DOSE: Energy per unit mass imparted by ionizing radiation to a small mass of material. *See also* rad.

ABSORPTION: Deposition of energy in a medium traversed by x-rays or other ionizing radiation. *See also* attenuation; x-ray attenuation.

ACCELERATOR (PARTICLE): A device that accelerates charged subatomic particles to very great energies. These particles may be used for direct medical irradiation, producing x-rays and neutrons, and for basic physical research. Medical units include linear accelerators, Van de Graaff units, betatrons, and cyclotrons.

AFTERLOADING: A brachytherapy technique which reduces operator hazard. Receptor tubes are first placed in the patient at leisure, and the radioactive sources then added *after* positions are verified to be acceptable.

Cycling: A particular afterloading technique employing a single high intensity source.

AIR-R: Exposure delivered directly from the photon source itself. *See also under* depth dose data terms and exposure.

AIR-R—IONIZATION CHAMBER MEASUREMENT:

Cable instrument: One employing a cable from the ionization chamber to its instrument. Its use provides both total dose (R) and dose rate (R/m) indications.

Condenser chamber instrument: Measures the total charge collected during a timed interval, from which *average* beam intensity during the interval may be obtained (R/m).

Field instrument: A robust, portable instrument used at clinical installations to measure beam intensity, HVL, and so forth (e.g., Victoreen or similar R-meter).

Roentgen (R): A special unit of exposure equal to 2.58×10^{-4} coulombs per kg of air (or the equivalent, 1 esu per 0.001293 g of air). See also exposure.

Types of chambers: There are two basic kinds of ionization chambers:

Cavity chamber: Primary beam traverses the charge-collecting electrodes enclosing the gas volume.

Free air chamber: Primary beam misses these electrodes.

ALPHA PARTICLE: Nucleus of a helium atom, consisting of two protons and two neutrons, ejected as a unit during disintegration of some radioactive nuclides (generally of high Z).

A-MODE DISPLAY: Direct pulse display of radar or ultrasound signals. See also under ultrasonography—technical.

ÅNGSTROM UNIT (Å): 10^{-8} or 0.000 000 01 centimeter.

ANNIHILATION: Destruction of positive and negative subatomic particles of similar mass by direct combination, producing corresponding energy release. See also under positron.

ARC SECTOR SCANNING: An ultrasound scanning technique in which the transducer is rocked as it moves about the patient. See also under ultrasonography—technical.

ASSAY: Determining the nature and strength of a radioactive, chemical, or pharmacologic preparation. See also radioactivity assay.

ATTENUATION: Removal of energy from a beam of ionizing radiation when it traverses matter, by deposition of energy there (absorption), and/or by deflection of energy out of the beam (deflection attenuation). See also x-ray attenuation.

AVERAGE LIFE: See under radioactive decay.

BACKGROUND: A term used in radioactivity measurement:

Count rate: That count rate produced by background radiation. The actually measured total rate must be reduced by the background rate to obtain that produced by the quantity being measured.

Radiation: That from all sources other than the one being

measured. In practice, these include other man-made ionizing radiation sources as well as naturally occurring radioactive sources and cosmic rays.

BACKSCATTER: Radiation deflected back through the plane of entry into an object. In orthovoltage x-ray therapy, one generally has reference to the skin location; in supervoltage x-ray therapy, to the depth of maximum ionization.

BARRIER: Short for radiation barrier or shield. *Primary:* one struck by the primary beam (from the source); *secondary:* one struck by the secondary beam. See also beam.

BARYTES: Natural barium sulphate. Sometimes used to increase the x-ray attenuation provided by plaster and concrete. Other high and intermediate atomic number materials are also used, such as litharge (PbO), steel, and of course lead.

BATCH DISPOSAL: Disposal of radioactive wastes in batches, on land and sea.

BEAM: The stream of ionizing radiation particles or photons, in general. In photon beams (x-ray and teletherapy machines) one speaks of:

Primary beam: The photon stream originating at the target or radioactive source.

Secondary beam: Photons originating in objects irradiated by the primary beam. These are primarily scattered photons. See also x-ray attenuation mechanisms.

Stray beam: Radiation other than that from the primary beam. In addition to secondary radiation, this includes direct or leakage radiation which comes directly from the target or source through the container in directions other than those desired.

BEAM BARRIER OR INTERCEPTOR: A large metal shield assembly used in some teletherapy machines to absorb almost all the primary beam as well as radiation scattered through as much as $\pm 30^\circ$ from the central ray. It is used to reduce the required thickness of walls struck by the primary beam.

BEAM FLATTENING FILTER: Special filter used in betatrons and linear accelerators, for x-ray beams, to obtain flat isodose curves. See also x-ray therapy—correctional filters.

BETA PARTICLE: Negative or positive electron ejected from a nucleus during some radioactive transformations. See also radioactivity particles.

BETATRON: An electron beam accelerator used to produce electron and x-ray beams with energies from 9 MeV through 45 MeV (medical units), by use of $\frac{1}{4}$ cycle of an ac magnetic field.

① Leakage
② Scatter

- BIOLOGIC HALF LIFE (T_b):** The time needed to clear half of the *non-radioactive* material from an organ in vivo. *See also* radioactive decay.
- B-MODE DISPLAY:** Display of pulse *peaks* only. *See also* ultrasonography—technical.
- BOLUS:**
In radiation therapy: Additional tissue equivalent material placed around curved and irregularly shaped anatomic areas of a patient to facilitate dosage calculation and assure more uniform dosage distribution.
In diagnostic studies: A term used to designate a volume of injected tracer or x-ray contrast medium which travels in the body without substantial dispersion during a study.
- BONE-SEEKER:** A toxic agent selectively deposited for substantial periods of time in bone.
- BONE-SPARING:** In connection with x-ray beams, the phenomenon of reduced radiation injury to bone for the same roentgen dose with higher energy photons (supervoltage and 2 or 3 mm copper HVL or harder orthovoltage beams).
- BRACHYTHERAPY:** Treatment at short distances with sealed radioactive sources. *See also* radioactivity therapy.
- BRAGG-GRAY RELATIONSHIP:** Relationship basic to radiation dosimetry. *See also* depth dose data measurement.
- BREMSSTRAHLEN:** Continuous spectrum x-rays produced when charged particles are slowed down by collision with nuclei. *See also* x-ray production.
- BROAD BEAM TRANSMISSION:** Transmission of a beam having substantial cross-section. *See also* x-ray transmission.
- B-SCAN:** Anatomic cross-sectional display scan. *See* ultrasonography—technical.
- CADMIUM SULPHIDE (CdS) DETECTOR:** A single CdS crystal whose resistance is reduced in response to irradiation. *See also* electrical radiation detectors.
- CASSETTE:** Light-tight film holder. *See also* x-ray radiography.
- CATHODE RAYS:** In vacuum and gas discharge tubes, beams of electrons traversing the tube from the negative (cathode) to the positive (anode) electrode.
- CAVITY IONIZATION CHAMBER:** One whose electrodes enclose the ionized gas volume, and are traversed by the measured beam. *See also* air-R—ionization chamber measurement.
- CENTRAL RAY, CENTRAL RAY DEPTH DOSE:** *See* depth dose data terms.

- CHARACTERISTIC RADIATION:** Photon radiation emitted when excited crystals, molecules, atoms, or nuclei return to or nearer to their ground state. Such photons have energies characteristic of the particle and energy level interval involved. Equivalent to "fluorescent" radiation.
- CHARGED PARTICLE EQUILIBRIUM:** Referring to supervoltage x-ray beams, a situation in which the beam first traverses sufficient mass per cm^2 of beam before reaching the given location to assure production of maximum ionization there. This concept arises in dosage measurement (exposure, rads near the skin), and in treatment planning generally (skin-sparing, and treatment through lung and other air-containing cavities).
- CHROMOSOMAL INJURY:** See radiation injury, cellular.
- CINERADIOGRAPHY:** Radiography employing a movie camera to photograph the output phosphor of an image intensifier tube.
- COEFFICIENTS OF ATTENUATION AND ABSORPTION:** See x-ray attenuation coefficient.
- COINCIDENCE COUNTING:** A method of radioactivity measurement employing two detectors responding to the same radiologic event. Registration is usually made to occur only when *both* detectors record a signal, as in positron emitter scanning and tube noise rejection in liquid scintillation counting.
- COLLIMATION:** The use of attenuating material such as lead to define the dimensions and direction of a beam of radiation. Applications include:
In vivo radioactivity measurement: Collimation limits detected photons primarily to those from a particular location of interest. Four special types are used: narrow vs. wide angle; single-hole vs. multiple-hole focused.
X-ray work: Collimation to limit the radiation field to the area of interest. Types: fixed and adjustable diaphragms; cones.
- COMPENSATING FILTER:** See x-ray therapy—correctional filters.
- COMPTON SCATTER EFFECT:** Mechanism of interaction of a photon and electron, with transfer of some photon energy to the electron. See also x-ray attenuation mechanisms.
- CONTACT THERAPY:** Very short TSD x-ray therapy (5 cm and less) used to treat very superficial lesions. See also x-ray therapy machines.
- CONTINUOUS WAVE ULTRASOUND:** Non-pulsed ultrasound, used diagnostically to measure blood flow. See also ultrasound acoustical terms.
- CONTRIBUTORY DEFLECTION:** In broad beams, deflection which

contributes to the beam intensity beyond the absorber. *See* x-ray transmission—broad beam.

CONTROLLED AREA: A defined area in which the occupational exposure of personnel is under the supervision of the Radiation Protection Supervisor (RPS).

COUNTING INSTRUMENTS—TYPES:

Digital printer: An attachment to a scaler which records the total counts during given time intervals (preset time), or the time for equal count totals (preset count). Fast units combine the accuracy of scalers with excellent speed of indication.

Ratemeter: An instrument which provides a direct indication of count rate, by use of special electric circuits. There is always some accompanying loss of accuracy.

Scaler: An instrument which totalizes the number of received pulses and displays the result appropriately. It is normally used with a timer, in two possible ways:

PRESET TIME: Records total counts during preset time, like 1, 2, or 5 minutes.

PRESET COUNTS: Instrument stops timer after preset number of counts is registered.

Object: To assure desired statistical accuracy (counts) with varying activity.

Cow: *See* radioactivity cow.

CRITICAL TISSUE OR ORGAN: That part of the body most susceptible to radiation damage under the conditions of irradiation. Example: bone marrow during whole body irradiation; thyroid gland following inorganic ^{131}I administration.

CUMULATIVE TIMER: A device which indicates elapsed time after the start of fluoroscopic examinations. Such timers may shut off the machine automatically after a preset time, and must be reset before x-rays may again be produced.

CURIE (Ci): 3.7×10^{10} disintegrations per second. *See also* radioactive decay.

CUTIE PIE: A portable ionization chamber survey meter. *See* survey meters.

CYCLING: *See* afterloading—cycling.

CYCLOTRON: An accelerator used to produce high energy protons, deuterons, and other relatively heavy charged particles. Energies of the order of 20 MeV to 100 MeV may be achieved in modified versions. Such particles may be used for basic physics research. They are sometimes used medically directly for experimental therapy, but more often to produce radionuclides and neutron beams.

- DAUGHTER:** A nuclide product of radioactive decay. *See also* radioactive series; radioactivity cow.
- DEAD MAN EXPOSURE SWITCH:** A switch so constructed that the circuit-closing contact is maintained only by continuous pressure on the switch by the operator.
- DECAY SCHEME:** A description of the emitted particles and their energies during a radioactive transition. *See also* radioactive disintegration.
- DEFLECTION ATTENUATION:** Removal of energy from a beam by deflection out of the beam, rather than by deposition in an absorber. *See also* x-ray attenuation; attenuation.
- DELTA RAYS:** Outer orbit electrons driven out of atoms with substantial energy, so they ionize other atoms in turn.
- DEPTH DOSE DATA—BASIC CORRECTIONS:** Corrections to published data required because such data are obtained from calculations or measurements made with simplifying assumptions. Almost all such corrections tend to be minimized when supervoltage beams are employed.
- Beam asymmetry:* Intensities on opposite sides of incident portal generally differ.
- Beam obliquity:* Standard data assume central ray is perpendicular to surface.
- Patient curvature:* Standard data assume flat surfaces only are present.
- Tissue inhomogeneity:* Tissue density and composition variations exist in patients, but are ignored in most data. Such variations affect attenuation and scatter, as well as electron equilibrium.
- DEPTH DOSE DATA—MEASUREMENT:**
- Bragg-Gray relationship:* An equation relating the absorbed dose (rads) at a particular location in a patient or phantom to the density of ionization (ion pairs per g) in a minute gas cavity at that same location. The relationship assumes uniform ion density in the vicinity of the point in question.
- Extrapolation chamber:* A special ionization chamber, originally designed by Failla, providing means to vary accurately the separation of parallel planar electrodes at will; it is used with appropriate phantoms to provide accurate measurements of central ray depth dose and backscatter.
- DEPTH DOSE DATA—TERMS:**
- Central ray:* The straight line passing through the center of the source and the center of the final beam-limiting diaphragm.

Central ray depth dose tables: Tables providing both percent depth dose and backscatter data along the central ray for various field size, beam quality, and SSD values.

Exposure:

- AIR-R:** Exposure to the surface from the incident beam.
- BACKSCATTER-R:** Exposure to the surface from x-rays scattered back by material beyond the surface in a patient or phantom.
- DEPTH-R:** Exposure within the irradiated object.
- EXIT-R:** Exposure at the exit surface of the irradiated object, usually taken for the central ray.
- SURFACE-R OR SKIN-R:** Total surface or skin exposure, due to the combination of air-R and backscatter-R contributions.

Isodose chart: A set of isodose curves, usually drawn for cardinal values of percent of maximum absorbed dose (i.e., 90, 80, 70, and so on), which represent the distribution of dose over a particular plane surface including the central ray within the irradiated object.

Isodose curve or contour: A line along which the absorbed dose is constant. For x-rays up to 400 kV, isodose surfaces or curves may alternatively be drawn as surfaces or curves of constant exposure.

Maximum or peak absorbed dose: The maximum value of the absorbed dose in the beam; usually refers to that found along the central ray.

Percent depth dose: Depth dose per hundred units maximum or other reference dose.

Scatter factor: The ratio of exposure at a given point in a phantom to the exposure at the same point with the phantom absent.

DIAGNOSTIC HOUSING: Enclosure for a diagnostic x-ray tube which meets the specification of less than 0.1 R per hour leakage at a meter distance from the tube target. *See also* housing.

DIRECT ACTION: Production of radiation injury by ions produced initially during irradiation. *See* radiation injury.

DISCHARGE: Electrical breakdown by ionization of a gaseous insulator. *See also* geiger-mueller (G-M) tubes.

DISCONTINUITY: *See* x-ray attenuation discontinuity.

DISINTEGRATION CONSTANT: A constant, characteristic of a radionuclide, which characterizes its decay rate for a given number of atoms present. *See also* radioactive decay.

DISTANCE UNITS: 1 centimeter (cm) = 10,000 microns (μ) = 100,000,000 angstroms (\AA).

DODGING: *See* x-ray image—contrast manipulation.

DOPPLER PRINCIPLE: The change in apparent frequency of a wave caused by relative motion of the source and observer. *See also* ultrasound—acoustical terms.

DOSE EQUIVALENT (DE): A quantity used in radiation protection work that expresses on a common scale, for all types of radiation, an estimate of the biologically effective irradiation of exposed persons. It is defined as the product of the absorbed dose in rads and certain modifying factors. For photon and electron radiation, the DE is numerically equivalent to the dose in rads and the exposure in roentgens, to usual required accuracy.

DYNAMIC ERRORS: Errors arising from finite measuring speed of radiation detectors and instruments. *See also* radioactivity measurement errors.

EFFECTIVE HALF LIFE T_{eff} : *See* radioactive decay—biologic half life.

EFFECTIVE PHOTON ENERGY E_{eff} : A numerical description of x-ray beam quality, simply related to the HVL. *See also* x-ray quality.

EFFICIENCY OF RADIATION DETECTOR: The fraction of entering particles actually detected.

EFFICIENCY OF X-RAY PRODUCTION: The fraction of the energy of the electrons striking the x-ray tube target which appears as x-rays. *See also* x-ray production.

EINSTEIN MASS-ENERGY RELATIONSHIP: $E = mc^2$. *See also* nuclear reactions.

ELECTRICAL RADIATION DETECTORS: Devices which can produce direct electrical signals when irradiated with ionizing radiation. Examples: p-n and p-i-n junction diodes; CdS crystals.

ELECTRON VOLT: An energy unit convenient in considering atomic interactions, equal to the energy acquired by an electron falling freely through a potential difference of one volt. *See also* energy.

ENERGIZATION BY IONIZING RADIATION: Acquisition by crystals, molecules, atoms, and nuclei of energy from ionizing radiation. The absorbed energy may be sufficient to fracture the recipient particle, or merely to raise its energy state without basic dislocation:

Excitation: Storage of energy which may later be released as one or more photons, but without loss of particles. Examples of such photons: crystals and molecules—infrared or visible light; atoms—visible and x-ray photons; nuclei—gamma rays.

- Ionization:* When enough energy is absorbed by an atom or molecule, an orbital electron may leave the particle completely. This is ionization.
- Nuclear disintegration:* When enough energy is absorbed from an incident projectile by a nucleus, a neutron or proton or even larger particle may be ejected.
- ENERGY:** The ability to overcome forces that bind and attract material particles. In radiology the most useful unit is the electron volt (eV), which is the energy acquired by an electron falling freely through a potential difference of one volt. Multiples: 1 keV = 1000 eV; 1 MeV = 1,000,000 eV.
- ENERGY LEVELS:** Locations in atoms or nuclei of electrons or nucleons, respectively, corresponding to different degrees of energization of the atom or nucleus.
- Excited state:* Higher than ground state energy levels of particles in atom or nucleus.
- Ground state:* The lowest energy level of particles in the atom or nucleus.
- Metastable state:* A semi-permanent arrangement in which excited nuclei of some nuclides release gamma photons some time after excitation occurs.
- ENERGY LEVEL DIAGRAMS:** Graphic representations of the energy levels of atoms and nuclei.
- ENLARGEMENT RADIOGRAPHY:** Direct production of an enlarged radiographic image. Also called "magnification." *See also* x-ray exposure of film.
- EQUILIBRIUM, CHARGED PARTICLE:** *See* charged particle equilibrium.
- EQUILIBRIUM, RADIOACTIVE:** *See* radioactive equilibrium.
- EQUIVALENT ABSORBER:** The thickness of lead or dense concrete (density 147 pounds per cubic foot) required to attenuate a given beam in a manner equivalent to the absorber in question.
- EQUIVALENT FILTER:** The thickness of standard filter material, such as aluminum, copper, or lead, required to attenuate a given beam in a manner equivalent to the filtration in question. Usually this concept is used with reference to filtration of x-ray tube windows and fluoroscopic panels.
- EQUIVALENT PENUMBRA:** A term used in enlargement radiography. *See* x-ray exposure of film—special methods.
- EXCITATION:** *See* energization by ionizing radiation.
- EXPONENTIAL RELATIONSHIP:** $y = e^{-x}$. *See* radioactive decay; x-ray attenuation.
- EXPOSURE:** A measure of the ionization produced in air by x-rays or gamma rays. It is the sum of electrical charges on all of the

ions of one sign produced in air, when all electrons liberated by photons in a volume element of air are completely stopped in air, divided by the mass of air in the volume element. The special unit of exposure is the roentgen (R).

EXTENSIVE IRRADIATION: Irradiation of substantial volumes of an organism, but not the entire body. *See also* irradiation, extensive.

EXTRAPOLATION IONIZATION CHAMBER: A specialized laboratory dosimeter for precise central ray depth dose measurement. *See also* depth dose data measurement.

FIELD SIZE: The geometric area irradiated by a given beam. *See also* x-ray beam.

FILM DENSITY: Darkening of a developed film. *See also* x-ray film—useful terms.

FILTER (SIMPLE, COMPOSITE): An attenuator inserted in the x-ray beam near the tube to modify the beam quality in a desired way. *See also* x-ray filtration.

FILTRATION: Attenuation of an x-ray beam prior to irradiation of the preparation. *See also* x-ray filtration.

FISSION: A reaction in which an external neutron interacts with a heavy nucleus; the latter splits into two parts of intermediate atomic mass, and releases more than one neutron, on the average. Under suitable circumstances the reaction is self-sustaining. Such a *chain reaction* takes place in a fission (atom) bomb and in a nuclear reactor.

FLUORESCENCE: Radiation released by ionized and excited atoms and excited nuclei when they return to ground state or other lower energy levels. *Examples:*

Gamma rays: May be released by some excited nuclei after nuclear reactions.

X-rays: Characteristic photons released when inner electron orbits are restored following ionization of atoms.

Visible light: Light released when outer orbit electron locations are refilled, as from fluoroscopic and intensifying screens after exposure to x-rays, and from NaI(Tl) crystals struck by gamma rays in scintillation detection of radioactivity.

FLUORESCENT RADIATION DETECTORS: Devices that release light when struck by ionizing radiation. *See also* luminescent radiation detectors, fluorescent.

FLUOROSCOPY: Use of x-rays to obtain an immediately visible image. *See also* x-ray fluoroscopy.

- FOCAL SPOT SIZE:** The apparent x-ray source area of an x-ray tube. *See also* x-ray tube.
- FORWARD EMISSION EFFECT:** The tendency, particularly for higher beam energies, for a maximum intensity in a produced x-ray beam to exist in the direction of the original electron beam. *See also* x-ray production.
- FRACTION:** In radiotherapy, a single radiation treatment in a series. *See also* irradiation—timing.
- FRACTION IRRADIATION:** That delivered in separate treatments or fractions on a daily or similar basis. *See also* irradiation timing.
- FREE AIR IONIZATION CHAMBER:** One in which the x-ray beam does not strike charge-collecting electrodes. *See also* air-R—ionization chamber measurement.
- FREE RADICALS:** Transient chemical combinations during and immediately after irradiation, usually occurring in water. *See also* radiation injury, indirect action.
- FRICKE DOSIMETER:** A chemical system for measuring absorbed dose by the oxidation in acid solution of Fe^{++} to Fe^{+++} . It is especially useful for dosage standardization of high energy super-voltage photon and electron beams.
- FUEL RODS:** Rods containing fissionable material, such as Pu^{239} or U^{235} . *See also* nuclear reactor essentials.
- FUSION:** A nuclear reaction in which very light nuclei coalesce to produce heavier nuclei, with release of neutrons and enormous amounts of energy. The reaction requires that interacting particles possess great kinetic energy initially to overcome their mutual electrical repulsion. Examples: hydrogen bomb; interior of stars.
- GAMMA RAY OR SPECIFIC GAMMA EMISSION CONSTANT:** The exposure rate in roentgens per hour one centimeter away in air from a tiny 1 mCi radioactive source.
- GAMMA RAYS:** *See* radioactivity particles.
- GEIGER-MUELLER (G-M) TUBES:** Gas filled tubes with two coaxial electrodes maintained with very high electrical fields, so any gas ionization results in electrical discharge.
- Discharge:* The situation in a G-M tube following initial ionization by an entering particle, characterized by a sustained breakdown of the gas insulation, and passage of tube current limited primarily by the associated external circuit.
- Plateau:* A range of G-M tube voltage for which the counting rate is substantially independent of minor voltage changes.

Quenching: Termination of the G-M tube discharge, usually by electrical or chemical methods.

Recovery time: The time required after initiation of discharge before the next interacting particle can be uniquely detected by the G-M tube and its instrument, or more generally by *any* radiation detecting system.

GEOMETRY: In radiation detection, a term used to refer to the spatial arrangement of the radiation source relative to the detector, which determines the fraction of emitted particles intercepted by the detector.

GRID: A device used in x-ray diagnosis to minimize scatter reaching radiation detectors.

Grid lines: Images of the lead slats in a grid. They are minimized in Bucky operation.

Grid radius: In focused grids, the distance between the grid and the line where planes of the lead grid slats intersect. Operation at this distance minimizes the width of grid lines.

Grid ratio: The width of lead slats divided by their separation in the grid array.

Grid types:

BUCKY—grids which move perpendicularly to the slats during exposures, to minimize grid lines.

CROSSED—two grids used with slats oriented perpendicularly, to improve scatter removal.

FOCUSED—with slats' planes intersecting in a common single line.

PARALLEL—with slats parallel to each other.

GRID-CONTROLLED X-RAY TUBE: A tube with a third electrode in the cathode assembly, used for precise electrical switching to permit very short but accurately timed x-ray exposures. *See also* x-ray tube.

HALF LIFE: The time required, neglecting statistical fluctuation, for half the atoms of a radionuclide to disintegrate. *See also* radioactive decay.

HALF VALUE LAYER: An index of photon beam radiation quality. It is the thickness of appropriate attenuating material which, when inserted in a narrow beam, reduces its intensity in half.

HAZARD, RADIATION: A situation in which it is possible for persons to receive more than the maximum permissible limit (MPD) dose equivalent (DE) for occupational exposure.

H-D CURVE: *See* x-ray film.

- HEEL EFFECT:** See x-ray production.
- HERTZ (Hz):** See ultrasound—acoustic terms.
- HOUSING:** The enclosure surrounding an x-ray tube or teletherapy source, which provides required electrical and radiation protection as well as physical support. Modern x-ray tube housings are both *shockproof* and *rayproof*.
- Diagnostic:* A housing with sufficient shielding to reduce radiation leakage levels to 0.1 R/hr or less at one meter distance from the source.
- Therapeutic:* A housing with sufficient shielding to reduce radiation leakage levels to 1.0 R/hr or less at one meter, or 0.1 percent of the useful beam central ray intensity, whichever is greater.
- HYPERBARIC OXYGEN RADIATION THERAPY:** Radiation therapy administered with the patient specially prepared by his breathing pure oxygen at up to three atmospheres absolute pressure for a suitable preparation time and during treatment. The object is to increase injury to normally hypoxic tumor cells which are otherwise selectively protected from radiation by their hypoxia.
- IMAGE BLUR:** See x-ray image quality.
- IMAGE CONTRAST:** The variations in intensity across an image which render anatomic or other information visible. See x-ray image quality; luminance contrast.
- IMAGE INTENSIFICATION:** Deriving a brighter fluorescent image in x-ray or gamma ray diagnostic procedures. Devices used include image intensifier tubes, the Marconi and Cinelix systems, and ciné and television adaptations of all three systems.
- IMAGE LATITUDE:** See x-ray image quality.
- IMAGE QUALITY:** See x-ray image quality.
- INDIRECT ACTION:** Production of radiation injury by radicals rather than by ions produced initially during irradiation. See also radiation injury.
- INTEGRAL DOSE:** The total energy from ionizing radiation absorbed in a specified region. (It is the integral with respect to mass of the absorbed dose throughout the region.) It is given in units of gram-rads or a multiple, megagram-rads (a million times larger), where one gram-rad is 100 ergs. It provides the radiotherapist with a numerical guide with which to estimate systemic trauma to the irradiated individual. Often also called "volume dose."
- INTENSIFYING SCREEN:** In radiographic cassettes, the fluorescent screens held in close contact with the x-ray film during radio-

graphic exposure. They "intensify" the action of x-rays by producing an additional photographic image by their fluorescent light, thereby greatly reducing the required exposure time and patient dosage.

INTENSITY: See x-ray intensity.

IONIZATION: Release of an electron from an atom by ionizing radiation. See also energization by ionizing radiation.

IONIZING RADIATION: X-rays, gamma rays, and certain particle beams. See also radiation—ionizing.

IRRADIATION—EXTENT:

Local: To a relatively small area of the body, with minor integral dose.

Extensive: To large area of the body, usually including much of torso and head.

Whole body: To the entire body, sparing no part.

IRRADIATION—TIMING:

Fractionated: Delivery of the total dose in parts or "fractions," usually equal to each other, in a daily or similar schedule, for an overall period of weeks.

Massive: Full delivery of the total dose in a single rapid exposure.

Overall treatment time: The total period of a course of treatments.

Protracted: Delivery of the total dose in a prolonged exposure lasting at least three days.

Session: A single therapeutic irradiation.

ISOCENTRIC MOUNTING: An arrangement making it possible to move a radiation source easily about a treatment site while maintaining a fixed distance and beam angulation. Used in certain teletherapy and linear accelerator x-ray machines.

ISODOSE CHART: A chart showing distribution of depth dose produced by irradiation. See also depth dose data terms.

ISOMERIC STATE: An excited state of nuclei of a radionuclide, having an observable half life. Example: technetium-99m (^{99m}Tc).

ISOMERIC TRANSITION: A transition between two isomeric states of a nucleus or from an isomeric state to the ground state. Example: $^{99m}\text{Tc} \rightarrow ^{99}\text{Tc} + \gamma \uparrow$.

ISORESPONSE CURVES: Curves describing locations of a point radioactive source below a particular collimator yielding the same detector response.

ISOTOPES: Nuclides of the same atomic number but different atomic mass. See also nuclide.

- KINESCOPE RADIOGRAPHY:** Obtaining still or motion pictures by photography of the television image of a subject being studied using x-rays.
- LABEL (TRACER):** A material whose fate in the body after administration can be followed by virtue of signals obtainable from the material. *See also* tracer study.
- LATENT IMAGE:** The undeveloped image in an exposed film. *See also* x-ray film.
- LATENT PERIOD:** The period following irradiation before appearance of gross radiation injury. *See also* radiation injury.
- LEAD GLASS:** Glass of high lead content which exhibits substantial x-ray attenuation. Used when one desires transparency combined with radiation shielding, as in fluoroscopic screen assemblies and windows of radiotherapy rooms.
- LEAD RUBBER:** A preparation of lead oxide or salt powder or other attenuating material uniformly and permanently distributed in a rubber or plastic base, in the form of flexible sheeting. Such fabric is used in protective aprons and gloves, as well as directly to mask out areas in the treatment field in radiotherapy, and in other shielding applications as a lead substitute.
- LEAKAGE:** Escape of dangerous radioactivity or ionizing radiation from protective containers. *See also* housing; radium sources.
- LINE FOCUS TUBE:** X-ray tube with a cathode design yielding a linear focal spot, almost universally used for roentgenology. *See also* x-ray tube.
- LINEAR ENERGY TRANSFER (LET):** The linear rate of energy loss (locally absorbed) by an ionizing particle traversing a material medium, expressed usually in keV/ μ .
- LOCAL IRRADIATION:** *See* irradiation—extent.
- LUMINANCE CONTRAST:** The fractional departure of the luminance of a particular image on a radiograph from that of surrounding areas, as observed on a viewbox.
- LUMINESCENT RADIATION DETECTORS:** Devices which produce visible light as a result of irradiation by ionizing radiation.
- Fluorescent:* Glow *immediately* when irradiated [e.g., CdWO₄, ZnS, NaI(Tl)].
- Photoluminescent:* Glow *later when irradiated with UV light* (special glass).
- Thermoluminescent:* Glow *later when heated* (e.g., LiF, MgF₂).

- MARINELLI-QUIMBY DOSAGE FORMULAS:** Used for calculating dosage from beta and gamma radiation delivered to tissues containing radioactive material in uniform concentration, and in volumes large compared with the range of beta rays employed.
- MASSIVE IRRADIATION:** Irradiation delivered at one time, in a brief period. *See also* irradiation timing.
- MAXIMUM OR PEAK DOSE:** Maximum absorbed dose along the central ray below the irradiated surface. *See also* depth dose data terms. *Also, in multiple portal or rotational therapy, the term maximum dose refers to the highest summated dose in the treated area.*
- MAXIMUM PERMISSIBLE BODY BURDEN:** That amount of activity of a given radionuclide which, fixed permanently in the human body, delivers *any* of the following maximum yearly doses:
1. 15 rems for most individual organs of the body.
 2. 30 rems when the critical organ is the thyroid or skin.
 3. 5 rems when the gonads or whole body is the critical organ.
 4. For bone-seekers, the RBE-corrected dose must not exceed that from 0.1 μg of radium and its daughters.
- MAXIMUM PERMISSIBLE CONCENTRATION (MPC):** Limits set on water and air concentrations of radionuclides, for 40 and 168 hours exposure per week, which yield maximum permissible body burden values and their corresponding organ dosages.
- MAXIMUM PERMISSIBLE DOSE EQUIVALENT (MPD):** For radiation protection purposes, the maximum dose equivalent (DE) that a person or specified parts thereof shall be allowed to receive in rems in a stated period of time. The DE is the number of rads times factors correcting for the given beam quality (quality factor QF), distribution (distribution factor DF), and so on.
- METASTABLE STATE:** A situation in which a nucleus remains excited for some time before it releases its gamma ray energy. *See also* energy levels.
- MILLICURIE HOURS:** A measure of the total number of radioactive disintegrations to which a patient is exposed during brachytherapy. *See also* radioactive decay.
- MODERATOR:** Low atomic number material which slows down fast neutrons. *See also* nuclear reactor essentials.
- MODIFIED PHOTON:** One reduced in energy by a Compton interaction. *See also* x-ray attenuation mechanisms.
- MODIFIED SCATTER:** Compton scatter. *See also* x-ray attenuation mechanisms.
- MONITORING:** Surveillance of people and areas to assure radiation safety. *See also* radiation safety monitoring.

MOTION BLUR: Blur in an image resulting from motion during the exposure of the film.

MOTTLE: The spotted appearance of images in x-ray diagnostic studies when detail resolution of the detector is adequate to show sufficiently fine details. In radiography without screens, the image may show graininess, attributable to both the granular nature of the silver image and the corpuscular (photon) nature of the x-ray beam. In image intensifier units operated at very high gain, a "snowy" image may be observed. Mottle is inherently a statistical phenomenon, and is hence benefited when more tiny crystal elements or photon elements are involved in production of the image.

MULTICHANNEL ANALYZER: A very elaborate spectrometer. *See also* pulse height discrimination.

MULTIPLE PORTAL RADIATION THERAPY: An irradiation technique for securing desired uniformity of tumor dosage in radiotherapy. *See also* x-ray therapy—field arrangements.

MUTATION: Alteration of the genetic characteristics of chromosomes. *See also* radiation injury—cellular.

NARROW BEAM TRANSMISSION: Very small field transmission. *See also* x-ray transmission.

NEUTRON-GAMMA (N, γ) REACTION: One in which a nucleus captures a neutron, releasing a gamma ray in the process. E.g., $\text{Co}^{59}(\text{n},\gamma)\text{Co}^{60}$; $\text{Au}^{197}(\text{n},\gamma)\text{Au}^{198}$.

NEUTRON-PROTON (N,P) REACTION: One in which a neutron replaces a proton from a nucleus, e.g., $\text{S}^{32}(\text{n},\text{p})\text{P}^{32}$.

NEUTRON BEAM SOURCES:

Accelerator: Speeds up light ions to great energies; these ions bombard appropriate targets, such as deuterium, releasing neutrons: $\text{H}^2(\text{d},\text{n})\text{He}^3$.

Fission: Neutrons are produced copiously by nuclear fission in nuclear reactors. A wide range of neutron energies exists, with many slow neutrons present.

Neutron howitzer: An arrangement for bringing boron-9 in close contact with an alpha particle source. Neutrons result from the reaction $\text{B}^9(\alpha,\text{n})\text{C}^{12}$. Paraffin or similar absorber may be used as a neutron shield and collimator.

NEUTRON SPEED:

Fast: In the range of 0.01 MeV to 10 MeV.

Slow or thermal: In approximate thermal equilibrium with the moderator (order of 0.025eV).

- NOISE—IMAGE:** In x-ray diagnosis, noise arises from two basic causes, artifacts and mottle. Artifacts obscure or otherwise mar the image; they are of technical origin. Mottle is of statistical origin, and is inherent in x-ray image formation. *See also* mottle.
- NOISE—PHOTOMULTIPLIER:** In scintillation detection, spurious electrical pulses produced by the photomultiplier cathode even when there is no scintillation or other light signal present. These signals are primarily of thermal origin, and may be reduced by refrigeration in liquid scintillation counting units, where very small light signals must be meaningfully processed. Newer tubes employ special cathode surfaces to minimize such noise.
- NONIONIZING RADIATION:** In this text, this term refers to electromagnetic wave radiation of lower photon energy (such as radio and infrared waves), sound waves of moderate intensity, and mechanical waves generally; as contrasted with x-rays and those from radioactive materials.
- NUCLEAR REACTIONS:** Reactions in which nuclear composition and/or energy levels are changed. The three most important to medical radiology are radioactivity, fission, and fusion. Energy is released in all three, often produced by the destruction of as much as 0.1 percent of the total mass of the involved particles, according to the *Einstein equation*, $E = mc^2$, where m is the mass destroyed and c is the velocity of light, with all three terms in appropriate units.
- NUCLEAR REACTOR ESSENTIALS:**
- Control rods:** Cadmium-containing rods which readily absorb and trap neutrons, thereby reducing the number available to interact with fuel atoms, and hence the reaction rate.
 - Cooling:** Heat must be removed to prevent damage to reactor components. This is accomplished using various cooling liquids pumped through the system; ultimately steam may be produced to generate electrical power.
 - Fuel rods:** Containers with fissionable materials like U^{235} or Pu^{239} .
 - Moderator:** Carbon or other low Z material which slows down fast neutrons to velocities at which they interact more readily with fuel atoms.
 - Shielding:** Shielding is required to protect persons in surrounding areas from both neutron and gamma radiation.
- NUCLIDE:** A given variety of nucleus, with its own atomic number-atomic mass combination.
- Isotopes:** Nuclides with the same atomic number (same chemical element) but different atomic mass numbers.

- Radioisotopes:* A widely used term, less preferred, for radionuclides.
- Radionuclide:* A radioactive nuclide.

- OCCUPANCY FACTOR (T):** The factor used in computing radiation protection barriers, by which workload must be multiplied to correct for the fact the area may sometimes be unoccupied when the source of ionizing radiation is turned on.
- OFF-FOCUS RADIATION:** Radiation from other than the focal spot. *See also* x-ray tube.
- ORTHOVOLTAGE:** X-rays generated at 140 through 400 kVp. *See also* x-ray therapy machines.
- OVERALL TREATMENT TIME:** Total period of a course of radiation therapy. *See also* irradiation—timing.
- OXYGEN EFFECT:** Dependence of cellular response to irradiation upon the oxygen tension. *See also* radiation injury—factors.

- PAIR PRODUCTION:** A mechanism of interaction with matter of very high energy photons (1.02 MeV and above) in which the photon vanishes, producing an ordinary and a positive electron. *See also* x-ray attenuation mechanisms.
- PARENT:** A radionuclide, whose decay product is referred to as a “daughter.” Used in connection with series of radionuclides, generally. *See also* radioactive series.

PARTICLE ACCELERATORS—BASIC EQUATIONS:

Relativity correction: An expression for a particle’s mass (m) at velocity v , relative to that at rest (m_0), and c , the velocity of light:

$$m = m_0 / \sqrt{1 - (v/c)^2}$$

Cyclotron resonance equation: An expression for the required frequency (f) of the voltage impressed across the dee system, in terms of the magnetic field strength (B) and the charge to mass ratio (e/m) of the accelerated particle:

$$f = (1/2\pi)(e/m) B$$

- PENUMBRA:** Indistinctness of the margin of a beam arising from the finite size of the source.
- PERCENT DEPTH DOSE:** Dose at depth per hundred units at a

reference location, usually the skin or maximum ionization depth. *See also* depth dose data—terms.

PERSONNEL MONITORING: Measurement of the dose equivalent (DE) received by personnel during both specific hazardous procedures and more prolonged work periods such as a month.

PHANTOM: A volume of tissue equivalent material used for dosimetry or evaluation of diagnostic techniques. In the former use, it is generally constructed large enough to provide adequate scatter; in the latter, to simulate the shape and attenuation of some part of the human body.

PHOSPHORESCENCE: The delayed appearance of fluorescent and gamma radiation, e.g., fluorescent screen afterglow during fluoroscopy, and isomeric transition (IT) radiation release in radionuclides (as in technetium-99m).

PHOTOELECTRIC EFFECT: A mechanism of interaction of photons with release of orbital electrons and complete disappearance of the photon. *See also* x-ray attenuation mechanisms.

PHOTOELECTRON: An electron released by the photoelectric effect. *See also* x-ray attenuation mechanisms.

PHOTOFLUOROGRAPHY OR PHOTOROENTGENOGRAPHY: An x-ray technique used in mass survey x-ray work. *See also* x-ray radiography.

PHOTOLUMINESCENT RADIATION DETECTORS: Certain glasses used for in vivo dosimetry. *See also* luminescent radiation detectors.

PHOTOMULTIPLIER TUBE: A special vacuum photoelectric cell containing its own built-in current amplifying means which greatly increases the current yield from a given light signal. Used in scintillation detection and phototiming apparatus.

PHOTON: A quantum or energy bundle in an electromagnetic wave which constitutes the unit of energy associated with a given wavelength. The energy is related to the wavelength by the relationship: $E = 12.4/\lambda$, where E is in KeV, λ , in Å.

PHOTOTIMING: An automatic diagnostic x-ray exposure system. *See also* x-ray exposure control—timing.

P-I-N JUNCTION DIODE: A p-n junction diode of special design to increase its sensitivity to both light and ionizing radiation.

PLATEAU: The optimum operating voltage range for G-M tubes. *See also* geiger-mueller (G-M) tubes.

P-N JUNCTION DIODE: A specially treated semiconductor crystal, usually Si, which generates a voltage or current when irradiated.

POSITRON: A particle of the same mass as an ordinary electron, but with a positive rather than negative unit charge.

Annihilation: Positrons tend to interact with ordinary elec-

trons, especially when their velocity is low. In the reaction both particles are destroyed, with the release of two annihilation photons of 0.51 MeV each.

Sources: Positrons are produced in beta (+) radioactive decay, and in pair and triplet production.

POTENTIATING AGENT: One which *increases* radiation injury from a given radiation dose. *See also* radiation injury factors.

PRIMARY BARRIER: A radiation barrier intercepting the primary beam. *See also* barrier.

PRIMARY BEAM: That originating at the ionizing radiation source. *See also* beam.

PRIMARY IONIZATION: That produced directly by a photon interaction. *See also* x-ray ionization.

PROBE SCINTILLATION COUNTER: One used for in vivo counting. *See also* scintillation detection.

PROCEDURAL ERROR: A type of systematic error. *See also* radioactivity measurement errors.

PROTECTED HOUSING: One shielded against radiation leakage to specified levels. *See also* housing.

PROTECTIVE AGENT: One which *reduces* the injury from a given radiation dose. *See also* radiation injury factors.

PROTRACTED IRRADIATION: Irradiation delivered over an extended time. *See* irradiation—timing.

PULSED ULTRASOUND: Ultrasound energy delivered for a few cycles only, in diagnostic studies. *See also* ultrasound—acoustical terms.

PULSE HEIGHT DISCRIMINATION: The use of special circuitry to differentiate scintillation pulses of different magnitudes, corresponding to absorbed photons of different energies in scintillation detection. *Terms involved:*

Channel: A discrimination unit set with a given sill and window (*see* below).

Multichannel analyzer: A versatile spectrometer which automatically records the distribution of photon energies reaching the scintillation crystal by use of up to hundreds of separate channels set with adjacent windows.

Sill: The lowest photon energy registered by a discriminator.

Spectrometer: An instrument with one or more discrimination channels.

Window: The range of photon energies accepted by the discriminator.

QUALIFIED EXPERT (QE): With reference to radiation protection, a person having the knowledge and training to measure ionizing

radiation, evaluate safety techniques, and advise regarding radiation protection needs. With reference to calibration of radiation therapy equipment, a person having, in addition to the above qualifications, training and experience in the clinical applications of radiation physics to radiation therapy.

QUALITY FACTOR: A factor in radiation protection work used to calculate the dose equivalent (DE) from the absorbed dose in rads, which takes into account the difference in biologic effect of the particular quality of radiation employed from that of 250 kV x-rays.

QUANTUM: See photon.

QUENCHING: Termination of a gaseous discharge, usually by chemical or electrical means. See also geiger-mueller (G-M) tube.

RAD: Unit of absorbed dose: 100 ergs per gram of absorber. See also absorbed dose.

RADIATION: Energy emanating from a point.

RADIATION INJURY:

Cellular: Very large ionizing radiation doses may kill the cell outright. More generally, *chromosomal* damage results. Gross changes may prevent continued cellular division, effectively *sterilizing* the cell. More localized chromosomal damage may alter the genetic nature of the cell, producing a *point mutation*, or simply *mutation*.

Direct action: Injury of vulnerable sites within cells by the radiologic lesion (see below).

Indirect action: Injury within cells produced by chemical agents (free radicals) resulting from the radiologic lesion, which can produce chemical lesions relatively remote from the initiating radiologic lesion.

Latent period: The time delay between exposure of tissue to irradiation and gross manifestation of the resultant changes. This delay may vary from a few weeks to 20 years or more, depending on the irradiation circumstances and the biologic end point.

Radiologic lesion: The ions and excited atoms resulting within a microsecond from the initial irradiation.

Recovery from radiation injury: The combined effect of intracellular recovery and tissue repair by cellular repopulation in irradiated tissue.

RADIATION INJURY—FACTORS:

Biologic: cellular radiosensitivity, previous trauma, circulatory status, and so forth.

Chemical: Agents may protect cells or potentiate the effect of radiation.

Protective agent: One reducing effectiveness of radiation (i.e., hypoxia).

Potentiating agent: One increasing the effect of radiation. One of the most important such agents is oxygen. Oxygenated cells are sterilized by lower doses than hypoxic cells. This *oxygen effect* serves to protect selectively some tumor cells which are hypoxic; this effect is presumed to be a factor in tumor recurrence.

Physical: Dosage distribution, timing of delivery, and type of radiation (LET), and so forth.

RADIATION INJURY—TYPES:

Reproductive: Injury to reproductive cells. The main concern in most radiation protection work is control of the mutation rate in human populations.

Somatic: Damage to other than reproductive cells, which affects primarily the irradiated individual.

RADIATION—IONIZING: Electromagnetic or particulate radiation capable of exciting and ionizing atoms, directly or indirectly, by interacting with matter.

RADIATION PROTECTION SUPERVISOR (RPS): The person at a radiation installation who is directly responsible for radiation safety.

RADIATION SAFETY MONITORING: Surveillance of radiation installations to assure radiation safety. Visual, area, and personnel monitoring are among methods employed.

RADIATION SOURCES—EXTERNAL AND INTERNAL: A distinction useful in practical application of MPD figures. External sources include x-ray, teletherapy, and other machines, as well as radioactive sources external to the body. Internal sources include all radioactive materials within the body. Both types of sources must be considered in evaluating the radiation exposure hazard to an individual.

RADIOACTIVE CONTAMINATION: A radioactive substance dispersed in materials or places where it is undesirable.

RADIOACTIVE DECAY: *See also* radioactive disintegration. Terms used in connection with concept:

Amount of activity: The rate of disintegration of nuclei of a radionuclide. Units: d/s, Ci, and multiples thereof.

Average life T_a : The average period of survival of atoms of a radionuclide: $T_a = 1/\lambda$ (λ defined below). It is 1.443 times the half life.

Biologic half life T_b : Many biologic processes involve transfer of material from an organ in a manner following an exponential curve. A term may then be defined, called the *biologic half life* T_b , which is the time required for half the *nonradioactive* material to be excreted from the organ in question. The actually observed disappearance of radioactive material from the organ reflects the combined effects of excretion and radioactive decay; the then measured or *effective half life* $T_{\text{eff}} = TT_b / (T + T_b)$.

Disintegration constant (λ): The fraction of the total number of atoms of a given radionuclide present at a given time which undergo radioactive transformation per unit time, neglecting statistical fluctuations.

Half life: The time required for half the atoms of a radionuclide to disintegrate, neglecting statistical fluctuations. $T = 0.693T_a = 0.693/\lambda$.

Millicurie hours: A measure of the total disintegrations (d) occurring during exposure of tissue to radioactivity. One mCi-hr is 13.32×10^{10} d, by definition of a mCi:

$$[\text{One mCi-hr} = (3.7 \times 10^7 \text{ d/s} \times (3600 \text{ s})) = 13.32 \times 10^{10} \text{ d.}]$$

Statistical nature: Radioactivity is a random process. This has two effects. First, decay is *exponential in character* (remaining activity vs. time is an exponential curve). Second, a given measurement of radioactivity will generally be higher or lower than that predicted by the decay curve; i.e., the activity *fluctuates*.

Units: One curie (Ci) = 3.7×10^{10} d/s; one mCi = 3.7×10^7 d/s; one μCi = 3.7×10^4 d/s.

RADIOACTIVE DECONTAMINATION: Reducing hazards associated with contamination to acceptable levels, by removing and/or immobilizing the material to prevent its spread.

RADIOACTIVE DISINTEGRATION OR RADIOACTIVITY: Spontaneous and inexorable transformation of nuclei of certain nuclides accompanied by loss of alpha, beta, and/or gamma rays.

Decay scheme: A diagram showing the types and relative probability of alternative modes of disintegration of a radionuclide. These diagrams may be quite simple, as with ^{14}C , or complex, as with ^{131}I .

Mechanisms: Alternative ways nuclei can undergo radioactive transformation. (α , β , EC, IT, and so forth.)

- RADIOACTIVE EQUILIBRIUM:** In a radioactive series, a situation in which the rates of decay of all daughter (product) nuclides of interest equal those of the parents.
- RADIOACTIVE SERIES:** A sequence of nuclides in which all are radioactive but the last, and in which each nuclide decays to produce the next nuclide in the series. Each nuclide is called the *parent* of the product, which in turn is called the *daughter*.
- RADIOACTIVITY:** See radioactive disintegration, activity.
- RADIOACTIVITY ASSAY:** Evaluation of the activity and nature of radioactive shipments and preparations.
- RADIOACTIVITY—CONCENTRATION BY TISSUES:** The selective deposition of radionuclides which is essential to useful external counting procedures, and of great interest in radiation protection. Mechanisms of regional localization include organ function; blood pools, both normal and abnormal; and mechanical trapping of particles in capillaries and bronchioles.
- RADIOACTIVITY COW:** An arrangement whereby a desired daughter radionuclide is obtained from a preparation of the parent, in form suitable for patient use. For example, Tc^{99m} as pertechnetate ion from Mo⁹⁹ as molybdenate ion.
- RADIOACTIVITY DETECTION—YIELD:** The ratio of detected counts to radioactive disintegrations (c/m vs d/m). It depends on at least three types of factors:
- Attenuation and scatter:* In the source material (self-attenuation), on the way, and while entering the detector.
 - Detector efficiency:* The fraction of entering photons or particles actually detected.
 - Geometry:* Many photons and other particles miss the detector entirely. Geometry refers to the spatial arrangement of the source and detector which determines this.
- RADIOACTIVITY DISPOSAL:** Safe disposition of radioactive wastes. Methods include batch disposal in drains, approved burial, AEC licensed incineration, and disposal by AEC licensed agents.
- RADIOACTIVITY MEASUREMENT ERRORS:** Two basic types exist:
- Statistical:* Arising from the random nature of radioactive decay. In general, there is fluctuation in repetitively measured values of the activity in a given sample. If the fluctuation arises from statistical causes only, there is a 67.45 percent chance any particular measurement of N counts will be within \sqrt{N} of this nominal value. This is called σ , the *standard deviation*. There is then a 95 percent chance of being within 2σ , a 99 percent chance of being within 2.6σ .
 - Systematic:* Arising from improper procedures and instru-

ment limitations. In addition to the more obvious types, errors can arise from *dynamic* factors. These include problems in measuring rapid physiologic changes and from use of radionuclides of very short half lives, which decay significantly during the period of a measurement.

RADIOACTIVITY PARTICLES: There are three types.

Alpha: Helium nuclei (2 protons + 2 neutrons).

Beta: Electrons, both plus and minus. *See also* beta particle.

Gamma: X-ray photons originating in excited product nuclei; nuclear characteristic radiation.

RADIOACTIVITY REFERENCE STANDARDS: Preparations of known radioactive strength and emitted particles.

RADIOACTIVITY THERAPY: Treatment of disease using radioactive preparations. There are three major types.

Brachytherapy: Treatment of disease by use of small sealed sources placed in and alongside lesions.

Solution therapy: Treatment of disease using radioactive solutions.

Teletherapy: Treatment at distances of about 20 to 100 cm by means of high activity gamma emitting radioactive sources in special shielded housings. The strength of such sources is specified in *RHM*, the exposure rate in *R/hr* at one meter, and *RMM*, the exposure rate in *R/m* at one meter.

RADIOGRAPHY: Obtaining an x-ray film image. *See also* x-ray radiography.

RADIOISOTOPE: *See* nuclide.

RADIOLOGIC LESION: The site in tissue of ionization and excitation from incident ionizing radiation. *See also* radiation injury.

RADIONUCLIDE: *See* nuclide.

RADIONUCLIDE PRODUCTION: Both particle accelerators and nuclear reactors are used.

Accelerators: Usually deuteron bombardment is used to transmute materials by (d,n) and (d,p) reactions.

Nuclear reactors: Fission fragments are often directly usable after chemical separation (^{131}I , ^{137}Cs , etc.). In addition, (n,p) and (n, γ) reactions produce radionuclides from irradiated materials.

RADIUM SAFE: A shielded enclosure for storage of radium sources not in use.

RADIUM SOURCES: Radium salt in a powder matrix is first sealed in cells; then these cells are in turn sealed into metal needles, tubes, and capsules. Needles are narrow and pointed in shape,

tubes slightly larger in diameter, and about 15 mm long. Capsules are essentially tubes with holes at one end for threading to facilitate withdrawal from the patient.

Filtration: Metal enclosure of the radium salts absorbs alpha and beta particles, which are undesirable in brachytherapy. In addition, the assembly is given mechanical strength and the dangerous contents sealed away from tissues.

Leakage: Leakage of both radon gas and radium salt is a potential hazard of all radium sources, and must be corrected immediately when found or the source discarded.

RATEMETER: An instrument giving an indication of dose or counting rate (R/m or c/m).

RAYPROOF: Term used to describe an adequately radiation shielded x-ray housing. *See also* housing.

RECIPROCITY LAW: *See* x-ray film.

RECOIL ELECTRON: Electron released in a Compton interaction. *See also* x-ray attenuation mechanisms.

RECOVERY: *See* radiation injury.

RECOVERY TIME: *See* geiger-mueller (G-M) tubes.

REFERENCE STANDARDS: *See* radioactivity reference standards.

RELATIVE BIOLOGIC EFFECTIVENESS (RBE): The relative biologic potency of a given kind of ionizing radiation. Quantitatively, the number of rads of 250 kV x-rays required to produce a given biologic end point divided by the number of rads of the given type of radiation needed to produce the same effect, for identical timing, dosage distribution, and other relevant exposure factors. Used primarily in radiobiologic research and radiotherapy; the more approximate *quality factor* (QF) is used in radiation protection work.

RELATIVITY CORRECTION: An equation expressing the mass of a fast moving particle as a function of its velocity. *See also* particle accelerators.

REM: A unit of MPD. *See also* maximum permissible dose equivalent.

RHM: R/hr at one meter of a teletherapy source.

RMM: R/m at one meter of a teletherapy source.

ROENTGEN (R): *See under* air-R-ionization chamber measurement; exposure.

ROENTGEN-RAD CONVERSION FACTOR (F): The number of rads of absorbed dose delivered per roentgen exposure. This depends on the absorber material and the photon spectrum. However, allowance must be made for any relevant departure from electron equilibrium. Tables are available (N.B.S. Handbook 87) for vari-

ous beam qualities and human tissues and other materials of clinical interest.

ROLL FILM CHANGER: A device used in circulatory studies to obtain sequential radiographs at a rate of up to six per second, by rapidly advancing the film *in between* successive exposures. The film is firmly clamped between two intensifying screens *during* exposures. Roll film changers are faster than cassette changers, and give finer details than cineradiography.

ROOT MEAN SQUARE (RMS) VOLTAGE: The value of a constant dc voltage yielding the same heating effect in a pure resistance as the voltage in question.

ROTATING ANODE: Anode design used in modern high rating x-ray tubes, in which the anode rotates to increase the effective target area. *See also* x-ray tubes.

ROTATIONAL THERAPY: Treatment in which the beam moves about the patient to increase total percent depth dose. *See also* x-ray therapy field arrangements.

SCALER: A totalizing counter, usually used with a timer. *See also* counting instruments.

SCANNING: Measurements of the *in vivo* spatial distribution of radioactivity in a given area.

Basic systems: Mechanical—with dot, photographic, and color displays. *Cameras:* Anger, Bender-Blau, image tube, and spark type.

Multiple detector systems: positron-coincidence; two-detector isoresponse; simultaneous multiple record; and body section scanning.

SCINTILLATION DETECTION: Detection of alpha, beta, gamma, and x-rays by use of fluorescent materials and photomultiplier tubes.

Detector system: Crystal (single or multiple) or liquid; photomultiplier tube; and preamplifier.

Type assemblies: Probe and well counters.

Type detectors: Alpha—ZnS screen; beta—scintillant liquid; gamma or x-ray—NaI(Tl) crystal.

SECONDARY BARRIER: Protection barrier against secondary beams. *See also* barrier.

SECONDARY BEAM: One other than that emerging usefully from the ionizing radiation source. *See also* beam.

SECONDARY IONIZATION: That produced by particles released in matter by photons. *See also* x-ray ionization.

SEED: A small radioactive source of short half life, usually less than 1 mm in diameter and 3 mm long, and most commonly

- employing radon or gold-198, used in brachytherapy. Usually it is left in place permanently.
- SEQUENTIAL RADIOGRAPHY:** See x-ray sequential radiography.
- SESSION:** An individual x-ray treatment. See also irradiation timing.
- SHALL AND SHOULD:** In NCRP radiation protection recommendations, *shall* means the requirement is necessary to meet currently accepted standards of protection. *Should* means the requirement is recommended or advisable, and is to be applied when practicable.
- SHIELDING:** The use of absorbers to reduce radiation intensity levels around sources of ionizing radiation in order to control radiation hazards.
- SHOCKPROOF CABLES:** Special cables for connecting x-ray tubes to their transformer units. Use of a grounded conducting outer electrode and thick high dielectric strength electrical insulation reduces shock hazards to negligible levels.
- SHOCKPROOF TUBE HOUSING:** One with negligible shock hazard. See also housing.
- SILL:** The minimum detectable pulse height of a discriminating amplifier, usually in a scintillation counting system. See also pulse height discrimination.
- SIMPLE SCATTER:** Thomson or unmodified scatter, in which photon energy remains unchanged. See also x-ray attenuation mechanisms.
- SINGLE PORTAL THERAPY:** Treatment to a single area of the patient. See also x-ray therapy field arrangements.
- SKIN-R OR SURFACE-R:** Exposure at skin surface. See also depth dose data terms.
- SKIN SPARING:** In supervoltage beam therapy, the reduced skin injury per roentgen exposure when electron equilibrium is not present at the advance portal.
- SOMATIC INJURY:** Injury to nonreproductive tissue. See also radiation injury.
- SOURCE STRENGTH:** In general, the activity of a radioactive source. Also relevant to brachytherapy is the active length of the source, which is the total length of radioactive materials in the source.
- SPECIFIC ACTIVITY:** Relating to solid or teletherapy sources, the activity per unit mass in Ci/g, kCi/g, μ Ci/g, and so forth. Relating to radioactive solutions, the activity per unit volume, in μ Ci/ml, and so forth.

- SPECIFIC GAMMA EMISSION CONSTANT (Γ):** The exposure rate in roentgens per hour at 1 cm from a tiny source of a gamma emitting radionuclide, without scatter or attenuation.
- SPECTROMETER:** An instrument which permits selection of sill and window in scintillation detection. *See also* pulse height discrimination.
- SPECTRUM:** Description of the photon energy composition of an x-ray beam. *See also* x-ray production.
- SPOT RADIOGRAPHS:** Those exposed during the progress of a fluoroscopic examinations. *See also* x-ray radiography.
- STANDARD DEVIATION (σ):** *See* radioactivity measurement errors.
- STATISTICAL UNCERTAINTY:** *See* radioactive decay; radioactivity measurement errors.
- STEREOSCOPY:** A special radiographic study yielding depth visualization. *See also* x-ray exposure of film—special methods.
- STERILIZATION (CELLULAR):** Loss of cellular reproductive capacity. *See also* radiation injury.
- STRAY RADIATION:** Ionizing radiation other than the useful beam. In medical radiology, primarily leakage and scattered radiation.
- SUBJECT CONTRAST:** That produced in the transmitted x-ray beam by the object traversed. *See also* x-ray image contrast manipulation.
- SUPERVOLTAGE PHOTON BEAM RADIOTHERAPY:** Radiotherapy using photon beams of effective energies 0.4 MeV and greater. Benefits over *orthovoltage* beam therapy include greater percent depth dose, simpler corrections of all sorts, and bone and skin sparing. Special filters are often used. *See also* x-ray therapy—correctional filters.
- SURVEY METERS:** Meters used to verify the radiation safety of an area, such as ionization chamber (cutie pie) and G-M counter units. They are generally battery powered for portable use.
- SYSTEMATIC ERRORS:** Those resulting from procedural errors and instrument limitations. *See also* radioactivity measurement errors.
- TARGET: TRANSMISSION AND REFLECTION:** Two basic types of x-ray target designs, used with high and low energy beams, respectively. *See also* x-ray tube.
- TELEETHERAPY:** Use of high activity gamma emitting sources in special housings as a substitute for supervoltage x-ray therapy machines. *See also* radioactivity therapy.

- TENTH VALUE LAYER:** In radiation protection barrier design, the thickness of absorber required to reduce the broad beam transmitted intensity by a factor of ten times.
- THERAPEUTIC HOUSING:** Enclosure for a therapeutic x-ray tube which meets the specification of less than 1.0 R/hr or 0.1 percent of central ray intensity at one meter distance from the tube target, whichever is greater. *See also* housing.
- THERMOGRAM:** An infrared scan, showing the skin temperature distribution over a surveyed area.
- THERMOGRAPHY:** The diagnosis of disease by measurement of patterns of skin temperature, using infrared detector scanning devices.
- THERMOLUMINESCENT DETECTOR:** A kind of solid state dosimeter. *See* luminescent radiation detectors.
- THOMSON SCATTER:** The same as simple scatter, in which there is no change in photon energy. *See also* x-ray attenuation mechanisms.
- TISSUE AIR RATIO (TAR):** The ratio of absorbed dose at a given point in a phantom to the absorbed dose which would be measured at the same location in free air within a volume of the phantom material just large enough to provide the maximum electronic buildup at the point of measurement.
- TOMOGRAPHY:** *See* x-ray exposure of film—special methods.
- TRACER STUDY:** A study in which the progressive movement of a chemical or bolus is followed during a process. In principle one may use any means which *labels* or *tags* the chemical or bolus to permit its identification, such as fluorescence, radiopacity, color, or other labels. Radioactivity offers the advantage of convenient in-vivo quantitative evaluation as well as identification; when it is the label, the study is a *radioactive tracer study*.
- TRANSMISSION CURVE, NARROW AND BROAD BEAM:** Curves of transmission vs. absorber thickness. *See also* x-ray transmission.
- TRUE ABSORPTION:** An obsolete term for x-ray absorption, in which energy of the attenuated beam is imparted to the absorber, as contrasted with *deflection attenuation*, in which energy is simply deflected out of the beam. The same photon interaction may involve both types of energy removal from the beam.
- TUBE RATING CHARTS:** *See* x-ray tube.
- ULTRASONOGRAPHY—TECHNICAL:**
- Arc sector scanning:* Scanning in which the transducer rocks back and forth in the scanning plane during its motion around the part.

Display modes: A-mode: pulse vs. time display; B-mode: pulse *height* vs. time display. In addition, there is B-scanning, which yields a two-dimensional representation similar to an anatomic cross-sectional view.

Scanning coupling means: Water bag, immersion, and contact scan methods exist.

ULTRASOUND—ACOUSTIC TERMS:

Ultrasound: Sound of frequency greater than 20,000 Hz. Typical use: 1 through 15 MHz.

CW ultrasound: Short for continuous wave ultrasound (many cycles), such as is used in Doppler flow measurements, as distinct from pulsed ultrasound, which is usually of 5 to 10 cycles duration.

Doppler principle: When a source of sound or other wave moves relative to an observer, the apparent frequency is raised or lowered in proportion to the relative velocity. The frequency change is then a measure of the relative speed.

Hertz, megahertz: One hertz (Hz) is one cycle per second; one megahertz, one million cycles per second.

Pulsed ultrasound: A burst of ultrasound, in practice, of the order of 5 or 10 cycles long, used for gauging distances by "sounding" or echo methods, in A and B display and B-scanning studies.

Reflection: The change in beam direction when it strikes an acoustic inhomogeneity. In ultrasonography, tissue stiffness and density differences are the main causes of such discontinuities, such as at the margins of cysts, bone, and soft tissue structures.

UNCONTROLLED AREA: One to which access is not limited for radiation hazard reasons. Also called "environs." *See also* controlled area.

UNMODIFIED SCATTER: Simple or Thomson scatter, in which the scattered photon energy is unchanged. *See also* x-ray attenuation mechanisms.

VOLTAGE SATURATION: In an ionization chamber, a mode of operation in which the charge collecting voltage is sufficient to gather up substantially all the ions produced by the measured radiation before they can recombine. The voltage required is called the *saturation voltage*. It must be increased for higher intensity beams.

- VOLUME DOSE:** The total energy delivered to an organism during irradiation with ionizing radiation. *See also* integral dose.
- WALL EFFECT:** Alteration of ionization chamber response to irradiation because x-rays strike the chamber walls. This causes variation in response to the same dosage with photon energy.
- WAVE FORMS—X-RAY MACHINE VOLTAGE:**
Types: Half wave, full wave, villard, constant potential.
Terms: peak; effective, or RMS, voltage.
- WEDGE ANGLE:** The angle through which isodose lines are shifted by a wedge filter. *See also* x-ray therapy—correctional filters.
- WELL SCINTILLATION COUNTER:** One designed for sample, rather than in vivo, counting. *See also* scintillation detection.
- WHOLE BODY COUNTING:** Evaluation of the amount and type of radioactivity in the entire body by external counting, usually of the scintillation type. Both liquid and crystal detectors have been used, with elaborate shielding to reduce background.
- WHOLE BODY IRRADIATION:** Irradiation of the entire organism, without exception. *See also* irradiation extent.
- WINDOW:** The range of pulse sizes accepted by a discriminator, as in a scintillation counting system. *See also* pulse height discrimination.
- WIPE TEST:** In radiation safety monitoring, measurement of removable contamination from a surface, such as a brachytherapy source, a teletherapy unit collimator, a laboratory bench, and so forth.
- WORKLOAD (W):** The degree of use of an x-ray or gamma ray source, used in computation of radiation protection barriers. For x-ray machines the workload is generally expressed in milliamperere minutes per week. For gamma ray beam therapy sources and for x-ray equipment operating at 400 pkV and above, the workload is usually stated in terms of the weekly exposure delivered by the useful beam at one meter from the source, and is expressed in roentgens or rads per week at one meter.
- X-RAY ABSORPTION:** Transfer of energy from an x-ray beam to atoms of irradiated material.
- X-RAY ATTENUATION:** Removal of energy from an x-ray beam traversing an object by absorption and/or deflection.
- X-RAY ATTENUATION COEFFICIENT:** A quantitative measure of the ability of an absorber to attenuate an x-ray beam, by both energy transfer and deflection. Formerly called "absorption coefficient."

Three terms are of interest:

Energy transfer coefficient: That portion of the linear or mass attenuation coefficient representing x-ray *absorption*.

Linear attenuation coefficient: The constant μ in the expression for the transmission T applicable to a narrow beam of monochromatic x-rays: $T = e^{-\mu x}$, where x is the absorber thickness and e the base of the natural logarithm system. μ is a measure of the total ability of the given absorber to remove energy of a particular photon energy from the beam. T is said to vary *exponentially* with x .

Mass attenuation coefficient: The linear attenuation coefficient μ divided by the absorber density ρ .

X-RAY ATTENUATION DISCONTINUITY: Referring to the log-log graph of μ/ρ vs. photon energy of a given element, the abrupt reduction in value of μ/ρ by the order of six times when the photon energy is reduced below the K-orbit binding energy. Applications include design of filters and diagnostic x-ray contrast media.

X-RAY ATTENUATION MECHANISMS: Ways in which photons can interact with matter.

Modified scatter (also called Compton and incoherent scatter): Photon deflection by electrons, accompanied by increase in wavelength λ according to the formula: $\Delta\lambda = 0.024(1 - \cos\theta)$ where θ is the angle through which the photon is deflected ($\Delta\lambda$ in Å). The scattered photon is reduced in energy, and is said to be *modified*. The lost photon energy appears almost completely in the form of kinetic energy of the free or outer orbit electron involved in the interaction (called a *recoil electron*).

Pair production: A process which may occur when photons of 1.02 MeV and greater energies interact with nuclei, especially those of heavier atoms. In this process the photon disappears completely. Its energy, however, reappears in two parts. The first part is as materialization of two electrons, one positive (a positron) and one negative (an ordinary electron). These are the *pair* which was formed. The second part is as kinetic energy of these particles, generally divided unequally between them. When the positron's velocity is reduced by collisions with atoms to moderate levels, it then undergoes a reaction with some nearby ordinary electron which is fatal to both particles: *annihilation*. Both particles disappear completely. They are replaced by two photons, each 0.51 MeV, which is exactly the energy equivalent to the mass of each annihilated particle. *Note:* annihila-

tion also occurs with positrons originating in other ways, as in β^+ radioactive disintegrations.

Photoelectric effect: A process in which the photon is completely absorbed in its interaction with an atom. Its energy is used to release an inner orbit electron (usually from the K-orbit), which emerges with all the original photon energy except that used to enable the electron to escape. The emerging electron is called a *photoelectron*.

Photonuclear reactions: Photons of sufficiently great energy (generally 15 MeV and greater) can dislodge particles from the nucleus, such as neutrons, protons, etc. These effects are of negligible practical effect in clinical radiology.

Simple scatter: Simple deflection of photons with insignificant energy delivered to the absorber; the deflected photon is thus altered in its direction. Also called *Thomson*, *coherent* and *unmodified* scatter.

X-RAY BEAM: The stream of x-ray photons emerging from the source and its collimator. *Terms:*

Central ray: The straight line passing through the center of the source and the center of the beam collimator opening.

Field size: A measure of the irradiated area size. There are two most useful conventions. The first is the *geometric* field size: the geometric projection on a plane perpendicular to the central ray of the distal end of the collimator as seen from the center of the front surface of the source. It is usually defined at the skin surface or the axis of rotation. The second is the *physical* field size, defined as the area included within the 50 percent of maximum dose isodose curve at the depth of maximum dose.

Penumbra: The region at the margin of a beam marked by a rapid falloff of intensity.

GEOMETRIC: The region of free space which is irradiated by photons coming from only a part of the source.

FREQUENT THERAPY CONVENTION: The separation in air of adjacent 20 percent and 80 percent of isodose lines, in a plane perpendicular to the central ray at the position of interest.

X-RAY DEFLECTION: Alteration of the direction of photons in an x-ray beam.

X-RAY DEFLECTION ATTENUATION: The resultant reduction in the transmitted intensity of the beam.

X-RAY EXPOSURE CONTROL—SWITCHING MEANS: These include relay type contactor switches, thyatron tubes, and grid controlled x-ray tubes.

X-RAY EXPOSURE CONTROL—TIMING:

Phototiming: A system, using a fluorescent screen with a photomultiplier tube, that terminates exposures automatically upon completion of the appropriate exposure.

Manual timing: Timers set manually by the operator on the basis of an estimate of the required time. Both synchronous and impulse or other electronic types are used.

X-RAY EXPOSURE OF FILM—SPECIAL METHODS: At least four special methods are used to improve diagnostic studies:

Enlargement radiography: Obtaining an x-ray image substantially larger than in ordinary radiography (2 or 3 times larger than the object itself). Penumbra is increased disproportionately, so useful enlargement or magnification radiography requires the use of tiny focal spot tubes (0.3 mm usually). The relevant measure of geometric blur is *equivalent penumbra*, defined as the ratio of penumbra (P) to enlargement ratio (M): $P/M = (1 - 1/M) F$. F is the focal spot size.

Multiple views: Obtaining several exposures of the same part from different projections.

Stereoscopy: A method of obtaining depth perception in radiography. First, two radiographs are exposed of the same part with the central ray shifted through the appropriate interpupillary viewing angle. The films are then examined simultaneously using a suitable stereoscopic optical viewing device.

Tomography: Radiography of a body section. Also called planigraphy and stratigraphy. Three basic types exist: plane through the patient axis; plane perpendicular to the patient axis; and polytome (shifting the cut during the exposure). Details in the section are highlighted by deliberately blurring details on either side of the section.

X-RAY FILM—USEFUL TERMS:

Density (D): Blackness of the developed film, as indicated by its transmission T of white viewbox light: $T = 10^{-D}$.

H-D curve: The graph of film density vs. the logarithm of exposure.

Latent image: The invisible chemical image produced by radiation, rendered visible by chemical development.

Reciprocity law: Relating to film response, this law states that the film density is independent of the exposure time, so long as the total exposure is constant. It applies when films are exposed to x-rays alone, but *not* when they are

exposed to light. Consequently, the law does *not* apply to film exposures using intensifying screens.

X-RAY FILTRATION: Attenuation of x-rays by material prior to their use for irradiation.

Filter: An added absorber inserted into the beam near the tube to achieve a desired total filtration. Filters can be *simple* (one material only) or *composite* (2 or more materials). In x-ray therapy special filters may also be used to change the distribution of intensity across the beam. *See also* x-ray therapy—correctional filters.

Inherent: Filtration within the tube, its housing, collimator, and associated structures.

X-RAY FLUOROSCOPY: X-ray examination using fluorescent screens to derive a visual image from the x-rays emerging from the patient.

X-RAY IMAGE: The pattern of variation of x-ray intensity across the x-ray beam in the radiation transmitted by the patient during an x-ray examination. This is converted by the detector into the *detector image* (on film, fluorescent screen, TV screen, and so forth). The latter is used to obtain the actual *visual image* observed by the radiologist.

X-RAY IMAGE CONTRAST MANIPULATION:

Dodging: The process of deriving an image of different contrast or contrast distribution from the original radiograph. This can be done by photographic, electronic, and color viewing methods.

Subject contrast: Contrast resulting from attenuation of the x-ray beam by the object traversed. For a given object, it also depends on the tube kV and filtration, scatter, and the use of any artificial contrast media.

X-RAY IMAGE QUALITY: The combination of image attributes which determines the diagnostic usefulness of the final image.

The following are involved:

Blur: Indistinctness of the margin of the image. This is a function of penumbra, motion, and inherent indistinctness of the margin of the object itself.

Contrast: The fractional increase or decrease in luminance of the image over that of its surroundings.

Detector resolution: A measure of the sharpness of the image produced by the detector of an x-ray image with negligible blur.

Latitude: Ability of a procedure to provide acceptable contrast of all desired images in a radiograph of body parts having greatly different x-ray attenuation.

Noise: Statistical fluctuations and spurious images. *See also* noise—image.

X-RAY INTENSITY: The measure of the rate of arrival to a given location of x-ray energy. Four different types of definitions are employed in radiology:

Absorbed dose rate: Rads per minute received at a point of interest.

Exposure rate: Roentgens per minute received at a point of interest.

Photon flux rate: The number of photons received per unit time per unit area perpendicularly to the beam at a point of interest.

Physical intensity: Energy per unit time received per unit area perpendicularly to the beam at the point of interest.

X-RAY IONIZATION—SPECIFIC DEFINITIONS:

Primary: Ionization produced directly by photons.

Secondary: Ionization initiated by photoelectrons, recoil electrons, electron pairs, or other particles liberated by primary ionization.

X-RAY MACHINES: Equipment which generates and controls voltages and currents to the x-ray tube.

Factors: Variables which affect the quantity (amount) and quality (type) of x-rays delivered during an exposure. They include kV, machine wave form, filter, mA, time, and SSD.

Safety—electric: To protect the machine, various overload switches and fuses are used; to protect patients and personnel, interlock switches as well as shockproof cables and housings are used.

Safety—radiation: The primary beam is restricted to the desired field by the use of *rayproof* housings of the *diagnostic* and *therapeutic* types and by collimating devices.

X-RAY PRODUCTION:

Bremsstrahlen: X-rays produced by the slowing down or braking of high speed charged particles.

Characteristic x-ray spectrum: One containing only certain discrete photon energies.

Continuous x-ray spectrum: One containing all photon energies between an upper and a lower limit. This type applies to Bremsstrahlen.

Efficiency: The fraction of the energy in the electron beam converted into x-rays at the target. This is roughly proportional to target atomic number, and rises even more rapidly with kV.

Forward emission effect: In x-ray tubes, the tendency to pro-

duce maximum x-ray intensity in the direction of the incident electron beam.

Heel effect: In x-ray tubes with reflection targets, a tendency for the intensity to be lower toward the anode than the cathode end of the tube.

Spectrum: A graph showing the relative intensities of photon components in a beam vs. photon energy.

X-RAY QUALITY: The special characteristics, and hence penetrating and other properties, of an x-ray beam, most accurately characterized by its spectrum. Other criteria approximate this in practice: HVL, E_{eff} , λ_{eff} . Penetrating beams are loosely described as *hard*, more easily absorbed beams as *soft*.

X-RAY RADIOGRAPHY: The recording of x-ray images on photographic film.

Cassette: A light-tight container used to hold the x-ray film during exposure, usually between two intensifying screens to reduce the required exposure time.

Photofluorography or photoroentgenography: Radiography using a special camera to photograph the image produced by the x-ray beam on a fluorescent screen.

Spot radiography: Radiography performed during the fluoroscopic examination, using the same x-ray tube as that employed for fluoroscopy.

X-RAY SEQUENTIAL RADIOGRAPHY: The obtaining of radiographs of the same body area in rapid sequence to observe physiologic function. Both fast cassette and roll film changers have been used, and photography or TV tape recording of the output phosphor image of image intensifier tubes are also commonly used.

X-RAY TARGETS:

Reflection: Targets inclined to the incident electron beam, and used so that the x-rays emerge from the same surface struck by the electron beam. Used in virtually all tubes operated at 400 kV and below.

Transmission: Targets struck on one side by the electron beam, with the useful x-ray beam emerging from the *opposite* side of the target. This is done in supervoltage x-ray generation to take advantage of the forward emission effect.

X-RAY THERAPY—CORRECTIONAL FILTERS:

Beam flattening: A roughly conically shaped filter used in betatrons and linear accelerator x-ray beams to flatten the isodose curves. Such filters must be used in the range of

4MeV and higher electron beam energies, to correct for the great forward emission effect.

Compensating: A nonuniform filter inserted in the treatment beam to compensate for nonuniformity of attenuation in a patient, in order to obtain more uniform dosage at the tumor level.

Wedge: A filter of triangular cross-section used to produce inclination of the isodose lines to their normal direction (called the *wedge angle*). The use of such filters permits design of customized isodose summations which deliver very uniform dosage to anatomic areas otherwise more difficult to treat properly.

X-RAY THERAPY—FIELD ARRANGEMENTS:

Multiple portal or field: Treatment delivered through several skin areas in order to achieve a required large dosage at depth without excessive injury to the skin and other superficial tissues.

Rotational therapy: Treatment delivered with the beam moving about the patient, but always pointed towards the tumor. This can be achieved either by rotating the patient with the machine fixed in position, or by moving the source about the patient. If the beam makes a full revolution, the procedure is called *360° or full rotation*; if motion is only through a partial revolution, the procedure is called *arc sector rotational therapy*.

X-RAY THERAPY MACHINES:

Contact: Employing very low tube kV and filtration, with SSD values of 5 cm or less.

Superficial: Employing tube voltages of 60 through 130 kV, with filtrations generally from 0.5 mm Al inherent up through 2 to 4 mm Al added filter.

Orthovoltage: Using tube voltages of 140 kV through 400 kV, with filters of 1 mm Cu + 1 mm Al up through Thoraeus III usually, occasionally up to Thoraeus V, and at SSD values of 50 cm generally, rarely up to 70 cm.

Supervoltage: Using x-ray sources operated at 1 MV and above, as well as radium, cesium-137 and cobalt-60 teletherapy units.

X-RAY TRANSMISSION: That fraction of a beam's energy not attenuated by an absorber. More quantitatively, the beam intensity at a given location with the absorber in place divided by the intensity at the same location with the absorber removed.

Broad beam transmission: Transmission of a beam which

has significant cross-section, so that substantial scattered radiation reaches the point of interest beyond the absorber. Such radiation contributes to the measured transmission and is referred to as *contributory deflection or scatter*.

Curve: A graph of transmission vs. absorber thickness. This is a straight line for monochromatic radiation when plotted on a semilogarithmic graph, and approximates this shape for most well-filtered beams.

Narrow beam: Transmission when the beam is very narrow in cross-section, so deflected incident radiation cannot readily reach a given point beyond the absorber.

X-RAY TUBE:

Focal size: The length of the side of the effective focus.

Focus—effective: The area at the target, measured perpendicularly to the central ray, from which the x-rays emerge.

Focus—target: The area of the target actually struck by the electron beam.

Grid controlled tube: An x-ray tube with a third electrode near the filament, whose potential relative to that of the filament may be varied to turn the x-rays on and off, for precision x-ray exposure timing.

Line focus tube: One in which the target focus is rectangular, so the effective focus is square. This is the design of most modern x-ray tubes.

Off-focus radiation: X-rays originating from areas in the tube other than the focal spot.

Rotating anode: In modern diagnostic x-ray tubes of high mA ratings, anodes specially constructed so the actual surface impacted by the electron beam is continually changed during the exposure; consequently, fresh anode surface is always presented to the electron beam, thereby greatly increasing the tube current rating for a given focal spot size.

Rating charts:

SINGLE EXPOSURE: Charts showing maximum permissible tube current plotted vs. exposure time, for various kV values. Charts apply only for the appropriate wave form and line frequency.

REPETITIVE AND PROLONGED EXPOSURES: Charts showing the cooling characteristics of the tube anode and housing, with *heat units* plotted vs. cooling time interval. *Number of heat units* = number of kVp \times number of mAs of an exposure.

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