10 External Radiation Safety

BASIC PRINCIPLES

Radiation safety practice is a special aspect of the control of environmental health hazards by engineering means. In the industrial environment, the usual procedure is first to try to eliminate the hazard. An example of this is the successive replacement of benzene as a degreasing solvent—first by carbon tetrachloride and later by trichloroethylene. If elimination of the hazard is not feasible, an attempt is made to isolate the hazard. If neither of these techniques is practical, then the hazard can usually be controlled by isolating the worker. The exact manner of application of these general principles to radiation safety practice depends on the individual situation. Radiation safety practice is divided between two principal categories: the safe use of sources of external radiation and prevention of personal contamination resulting from inhaled, ingested, or tactilely transmitted radioactivity.

External radiation originates in X-ray machines and other devices specifically designed to produce radiation; in devices in which production of X-rays is a side effect, as in the case of the electron microscope; and in radioisotopes. If it is not feasible to do away with the radiation source, then exposure of personnel to external radiation must be controlled by concurrent application of one or more of the following three techniques:

1. Minimizing exposure time.
2. Maximizing distance from the radiation source.
3. Shielding the radiation source.

Time

Although many biological effects of radiation are dependent on dose rate, it may be assumed, for purposes of environmental control, that the reciprocity relationship,

$$\text{dose rate} \cdot \text{exposure time} = \text{total dose},$$

is valid. For total dose that falls within one or two orders of magnitude of the radiation protection guide value, we have no data, neither clinical nor experimental, to contradict this assumption. Thus, if work must be performed in a relatively high radiation field, such as the repair of a cyclotron made radioactive by the absorption of neutrons or manipulation of a radiographic source in a complex casting, then the restriction of exposure time—so that the product of dose rate and exposure time does not exceed the maximum allowable total dose—permits the work to be done in accordance with radiation safety criteria. For example, in the case of a radiographer who must make radiographs 5 days per week while working in a radiation field of 0.25 mSv/h (25 mrems/h), overexposure can be prevented by limiting his daily working time in the radiation field to 48 minutes. His total daily dose would then be only 0.2 mSv (20 mrems). If the volume of work requires a longer exposure, then either another radiographer must be used or the operation must be redesigned in order to decrease the intensity of the radiation field in which the radiographer must work.

Distance

Point Source

Intuitively, it is clear that radiation exposure decreases with increasing distance from a radiation source. When translated into quantitative terms, this fact becomes a powerful tool in radiation safety. We will consider three cases: a point source, a line source, and a surface source.

In the case of a point source, the variation of dose rate with distance is given simply by the inverse square law, Eq. (9.22). Cobalt-60, for example, which emits one photon of 1.17 MeV and one photon of 1.33 MeV per disintegration, has a source strength that is approximated by Eq. (6.16b):

$$\begin{aligned}\Gamma &= 1.24\times10^{-7}\sum f_i \cdot E_i \frac{\mathrm{Sv\cdot m^2}}{\mathrm{MBq\cdot h}}\\ &= 1.24\times10^{-7}(1\cdot 1.17 + 1\cdot 1.33)\\ &= 3.1\times10^{-7}\frac{\mathrm{Sv\cdot m^2}}{\mathrm{MBq\cdot h}}\left(1.3\frac{\mathrm{R\cdot m^2}}{\mathrm{Ci\cdot h}}\right).\end{aligned}$$

For a 3700-MBq (100-mCi) source, the exposure rate at a distance of 1 m is about 1.15×10^{-3} Sv/h (115 mrems/h). If a radiographer were to manipulate this source for 1 hour per day, his maximum dose rate should not exceed 0.2 mSv/h (20 mrems/h) in order to stay within the weekly maximum guide of 1 mSv (100 mrems). This restriction could be attained through the use of a remote handling device whose length, as calculated from the inverse square law, Eq. (9.22), is at least 2.5 m.

If the radiography is to be done at one end of the shop, which is set aside exclusively for this purpose, then either a barricade must be erected outside of which the dose rate does not exceed the maximum allowable weekly rate, or, if this is not possible because of space limitations, a shield must be erected. If the barricade is used, its distance from the source must be such that the dose rate will not exceed

$$\frac{1\ \dfrac{\mathrm{mSv}}{\mathrm{wk}}}{40\ \dfrac{\mathrm{h}}{\mathrm{wk}}} = 0.025\ \frac{\mathrm{mSv}}{\mathrm{h}} \quad \left(2.5\ \frac{\mathrm{mrems}}{\mathrm{h}}\right).$$

By the inverse square law, this distance is found to be 2.3 m. Because of the inverse square effect, the dose rate increases rapidly as a person approaches a point source, and it decreases rapidly while the person moves away from the source. At any distance d from a point source of activity A and specific gamma-ray constant Γ, the dose-equivalent rate is given by

$$\dot{H} = \frac{\Gamma A}{d^2} \cdot w_{\mathrm{R}}. \tag{10.1}$$

The total dose to a person approaching a source during the time t is given by

$$H = \int_0^t \dot{H}\, \mathrm{d}t. \tag{10.2}$$

Equation (10.2) can be solved in terms of the velocity of approach v and the closest distance d_0 to which the source is approached. If we let

$$d = d_0 + vt, \tag{10.3}$$

then

$$\mathrm{d}d = v\ \mathrm{d}t, \tag{10.4a}$$

$$\mathrm{d}t = \frac{\mathrm{d}d}{v}. \tag{10.4b}$$

If we now substitute Eqs. (10.1) and (10.4b) into Eq. (10.2), we have

$$H = \int_{d_0}^{\infty} \frac{\Gamma \cdot A \cdot w_{\mathrm{R}}}{d^2} \cdot \frac{\mathrm{d}d}{v} = \frac{\Gamma \cdot A \cdot w_{\mathrm{R}}}{v} \int_{d_0}^{\infty} \frac{\mathrm{d}d}{d^2}. \tag{10.5}$$

Therefore,

$$H = \frac{\Gamma \cdot A \cdot w_{\mathrm{R}}}{v \cdot d_0}. \tag{10.6}$$

Line Source

In the case of a line source of radiation, such as a pipe carrying contaminated liquid waste, the variation of dose rate with distance is somewhat more complex mathematically than in the case of the point source. If the linear concentration of activity in the line is C_l MBq or curies per unit length of a gamma emitter whose source strength is Γ, then the dose rate at point p (Fig. 10-1), at a distance h from the infinitesimal length dl, is given by

$$\mathrm{d}\dot{D}_\mathrm{p} = \frac{\Gamma \cdot C_l \cdot \mathrm{d}l}{l^2 + h^2}, \tag{10.7}$$

and for the dose rate due to the activity in the total length of pipe, we have

$$\dot{D}_\mathrm{p} = \Gamma \cdot C_l \int_0^{l_1} \frac{\mathrm{d}l}{l^2 + h^2} + \Gamma \cdot C_l \int_0^{l_2} \frac{\mathrm{d}l}{l^2 + h^2}$$

$$\dot{D}_\mathrm{p} = \frac{\Gamma \cdot C_l}{h} \left(\tan^{-1} \frac{l_1}{h} + \tan^{-1} \frac{l_2}{h} \right)$$

$$\dot{D}_\mathrm{p} = \frac{\Gamma \cdot C_l \cdot \theta}{h}. \tag{10.8}$$

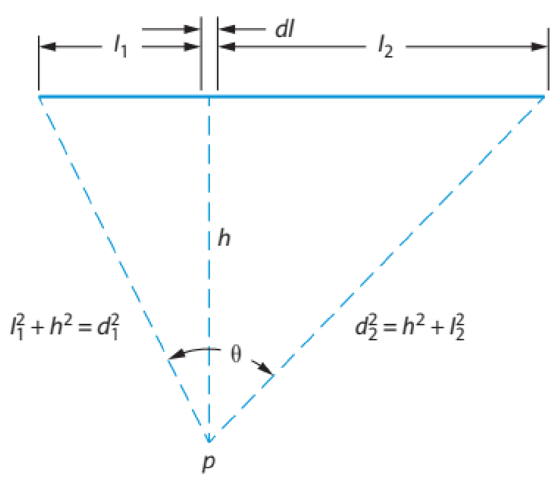

Figure 10-1. Geometry for computing gamma-ray dose rate at a finite distance h from a line source of uniform activity C_l per unit length.

Area (Plane) Source

Frequently the health physicist may find it necessary to know the quantitative relationship between dose rate and distance from a plane radiation source. If we have a thin source of radius r meters (Fig. 10-3) and a surface concentration C_a MBq/m^2 of a gamma emitter whose source strength is Γ Sv (air kerma) per hour per MBq at 1 m, then, the dose-equivalent rate at a point p, at a distance h along the central axis, is given by

$$\dot{H} = \int_0^r \frac{\Gamma \dfrac{\text{mSv} \cdot \text{m}^2}{\text{MBq} \cdot \text{h}} \cdot C_a \dfrac{\text{MBq}}{\text{m}^2} \cdot 2\pi r \, dr}{r^2 + h^2}$$

$$\dot{H} = \Gamma \cdot C_a \cdot \pi \cdot \ln \frac{r^2 + h^2}{h^2} \quad \frac{\text{Gy}}{\text{h}}. \tag{10.11}$$

If activity is given in Ci/m^2 and Γ in R-m^2/Ci-h, and since an exposure of 1 R corresponds to a dose equivalent of 1 rem (for safety purposes), the dose-equivalent rate in rems/h can be calculated with Eq. (10.11) by using the traditional units instead of the SI units.

$$\dot{H} = \Gamma \frac{\text{rem} \cdot \text{m}^2}{\text{Ci} \cdot \text{h}} \cdot C_a \frac{\text{Ci}}{\text{m}^2} \cdot \pi \cdot \ln \frac{r^2 + h^2}{h^2} \quad \frac{\text{rem}}{\text{h}}. \tag{10.12}$$

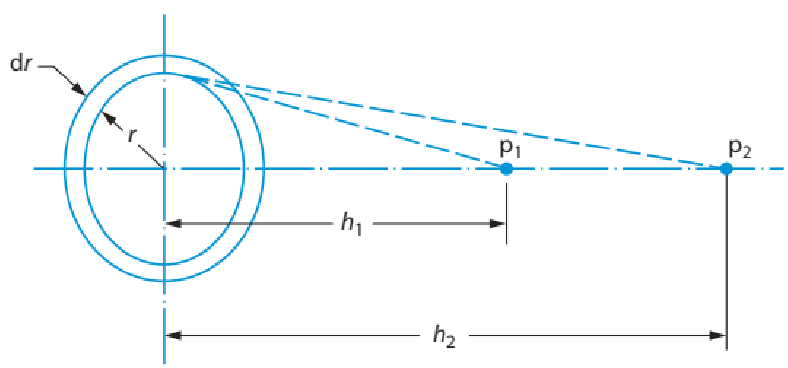

Figure 10-3. Geometry for calculating the variation of dose with distance from a plane source of radiation.

The ratio of the dose-equivalent rate at a distance h to the dose-equivalent rate at any other distance is given by

$$\frac{\dot{H}_1}{\dot{H}_2} = \frac{\ln\left[\dfrac{(r^2 + h_1^2)}{h_1^2}\right]}{\ln\left[\dfrac{(r^2 + h_2^2)}{h_2^2}\right]}. \tag{10.13}$$

Volume Source

In an infinitely large volume containing a uniformly distributed radioisotope, the energy emitted per unit volume is equal to the energy absorbed per unit volume. For example, consider the case of a 30-gallon metal drum, 50 cm in diameter and 79 cm high, containing aqueous ^{137}Cs waste at a concentration of 0.1 μCi/cm^3 (3700 MBq/m^3). Cesium-137 emits a 0.661-MeV gamma in 85% of its transformations. Since the attenuation coefficient for 0.661-MeV gammas in water is about 0.0896 cm^{-1} (Appendix E), an assumption of an infinite volume is reasonable. The estimated gamma-ray dose rate in the center of the drum is given by

$$\dot{H} = \frac{C\dfrac{\mu\text{Ci}}{\text{cm}^3}\cdot 3.7\times 10^4\dfrac{\left(\dfrac{\text{trans}}{\text{s}}\right)}{\mu\text{Ci}}\cdot \overline{E}\,\dfrac{\text{MeV}}{\text{trans}}\cdot 1.6\times 10^{-6}\dfrac{\text{erg}}{\text{MeV}}\cdot 3600\dfrac{\text{s}}{\text{h}}}{\rho\dfrac{\text{g}}{\text{cm}^3}\cdot 100\dfrac{\text{ergs/g}}{\text{rad}}}\cdot 1\frac{\text{rem}}{\text{rad}_\gamma}$$

$$\dot{H} = \frac{0.1\dfrac{\mu\text{Ci}}{\text{cm}^3}\cdot 3.7\times 10^4\dfrac{\left(\dfrac{\text{trans}}{\text{s}}\right)}{\mu\text{Ci}}\cdot \left(0.85\cdot 0.661\dfrac{\text{MeV}}{\text{trans}}\right)\cdot 1.6\times 10^{-6}\dfrac{\text{erg}}{\text{MeV}}\cdot 3600\dfrac{\text{s}}{\text{h}}}{1\dfrac{\text{g}}{\text{cm}^3}\times 100\dfrac{\text{ergs/g}}{\text{rad}}}\cdot 1\frac{\text{rem}}{\text{rad}}$$

$$\dot{H} = 0.12\frac{\text{rem}}{\text{h}} = 120\frac{\text{mrems}}{\text{h}}\left(1.2\frac{\text{mSv}}{\text{h}}\right).$$

The dose rate at the surface is 1/2 that in the center of the infinite mass, since the surface is irradiated from one side only, that is, it experiences 2π radiation rather than 4π radiation when it is completely surrounded by radioactivity. The surface gamma-ray dose rate, therefore, is $0.5 \cdot 120 = 60$ mrems/h (0.6 mSv/h).

The radiation doses rate from a source that is substantially less than infinitely thick, containing a uniformly distributed gamma-emitting isotope, may be estimated from the effective surface activity after allowing for self-absorption within the slab. Consider a large slab of thickness x m (Fig. 10-4), containing C_v MBq/m^3 of uniformly distributed radioactivity.

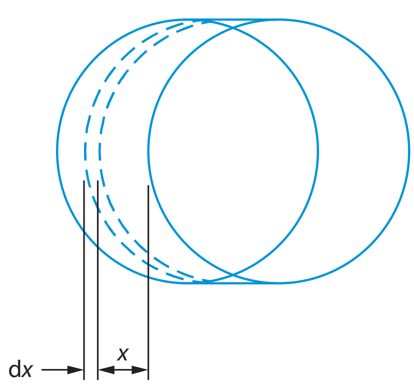

Figure 10-4. Conditions for setting up Eq. (10.14).

The linear absorption coefficient of the slab material is μ. The activity on the surface due to the radioactivity in the layer dx at a depth of x is

$$d(C_a) = C_v \, d\, x e^{-\mu x}. \tag{10.14}$$

Integrating Eq. (10.14) over the total thickness x yields the effective surface activity:

$$C_a = \int_0^t C_v e^{-\mu x} dx = \frac{C_v}{\mu}(1 - e^{-\mu x}). \tag{10.15}$$

Substituting Eq. (10.15) into Eq. (10.11) yields

$$\dot{H} = \pi \Gamma \frac{\text{mSv} \cdot \text{m}^2}{\text{MBq} \cdot \text{h}} \cdot \frac{C_v \dfrac{\text{MBq}}{\text{m}^3}}{\mu \text{ m}^{-1}}(1 - e^{-\mu x}) \cdot \ln \frac{r^2 + h^2}{h^2} \frac{\text{mSv}}{\text{h}}. \tag{10.16}$$

In traditional units, where activity is in curies and Γ is in units of rems per Ci per hour at 1 m, Eq. (10.16) becomes

$$\dot{H} = \pi \Gamma \frac{\text{rem} \cdot \text{m}^2}{\text{Ci} \cdot \text{h}} \cdot \frac{C_v \dfrac{\text{Ci}}{\text{m}^3}}{\mu \text{ m}^{-1}} \cdot (1 - e^{-\mu x}) \cdot \ln \frac{r^2 + h^2}{h^2} \frac{\text{rems}}{\text{h}}. \tag{10.17}$$

In the illustration above where the surface dose rate of the concrete in the 30-gallon drum was calculated as 60 mrems/h, let us calculate the dose rate on the central axis of the cylinder at a height of 30.5 cm (1 ft) above the top using Eq. (10.17). The absorption coefficient for 0.661-MeV gammas in water is 0.03284 cm^2/g (Appendix F), and the specific gamma-ray constant $\Gamma = 0.343$ R-m^2/Ci-h with $f = 0.962$ (Table 6-1). Substituting the respective values into Eq. (10.17), we have

$$\dot{H} = \pi \cdot 0.343 \frac{\text{R} \cdot \text{m}^2}{\text{Ci} \cdot \text{h}} \cdot \frac{0.1 \dfrac{\text{Ci}}{\text{m}^3}}{3.284 \text{ m}^{-1}} \cdot \left(1 - e^{-3.284\,\text{m}^{-1} \cdot 0.79 \text{ m}}\right) \cdot \ln \frac{(0.25 \text{ m})^2 + (0.305 \text{ m})^2}{(0.305 \text{ m})^2} \cdot 0.962 \frac{\text{rem}}{\text{R}}$$

$$\dot{H} = 1.5 \times 10^{-2} \text{ rem/h} = 15 \text{ mrem/h} (0.15 \text{ mSv/h}).$$

Shielding

In Chapter 5, Eq. (5.27), we saw that, under conditions of good geometry, the attenuation of a beam of gamma radiation is given by

$$I = I_0 e^{-\mu x}.$$

However, under conditions of poor geometry, that is, for a *broad beam* or for a very thick shield, Eq. (5.27) underestimates the required shield thickness because it assumes that

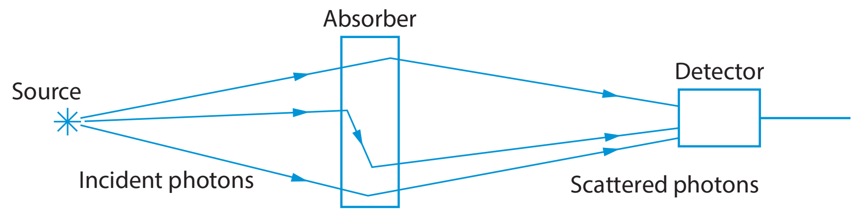

Figure 10-5. Gamma-ray absorption under conditions of *broad-beam* geometry, showing the effect of photons scattered into the detector.

every photon that interacts with the shield will be removed from the beam and thus will not be available for counting by the detector. Under conditions of broad-beam geometry (Fig. 10-5), this assumption is not valid; a significant number of photons may be scattered by the shield into the detector, or photons that had been scattered out of the beam may be scattered back in after a second collision. This effect may be illustrated by Figure 10-6, which shows the broad-beam and narrow-beam attenuation of ^{60}Co gamma rays by concrete. According to Eq. (5.27), about 7 in. (17.8 cm) of concrete shielding is required to transmit 10% of the incident ^{60}Co radiation, under conditions of good geometry. For a broad beam, however, Figure 10-6 shows that this thickness of concrete will transmit about 25% of the radiation incident on it. To transmit only 10% of a broad beam requires about 11 in. (30 cm) of concrete. In designing a shield against a broad beam of radiation, experimentally determined shielding data for the radiation in question should be used whenever they are available. (Broad-beam attenuation curves for radium, ^{60}Co, and ^{137}Cs for concrete, iron, and lead are given in Figures 10-6, 10-7, and 10-8.) When such data are not available, a shield thickness for conditions of poor geometry may be estimated by modification of Eq. (5.27) through the use of a *buildup factor B*:

$$I = B \cdot I_0 e^{-\mu x}. \tag{10.18}$$

The buildup factor, which is always greater than 1, may be defined as the *ratio of the intensity of the radiation, including both the primary and scattered radiation, at any point in a beam, to the intensity of the primary radiation only at that point.*

$$B = \frac{\text{primary + scattered radiation}}{\text{primary radiation}}$$

If multiple materials and buildup factors are necessary, Eq. (10.18) changes to

$$I = B_1 \cdot B_2 \cdot B_i \cdot I_0 e^{-(\mu_1 x_1 + \mu_2 x_2 + \mu_i x_i)}. \tag{10.19}$$

Buildup factors may apply either to radiation flux or to radiation dose. Buildup factors have been calculated for various gamma-ray energies and for various absorbers.[1] Some of

[1]ANS-6.4.3-1991, W2001. "Gamma-Ray Attenuation Coefficients and Buildup Factors for Engineering Materials," American National Standard, 1991.

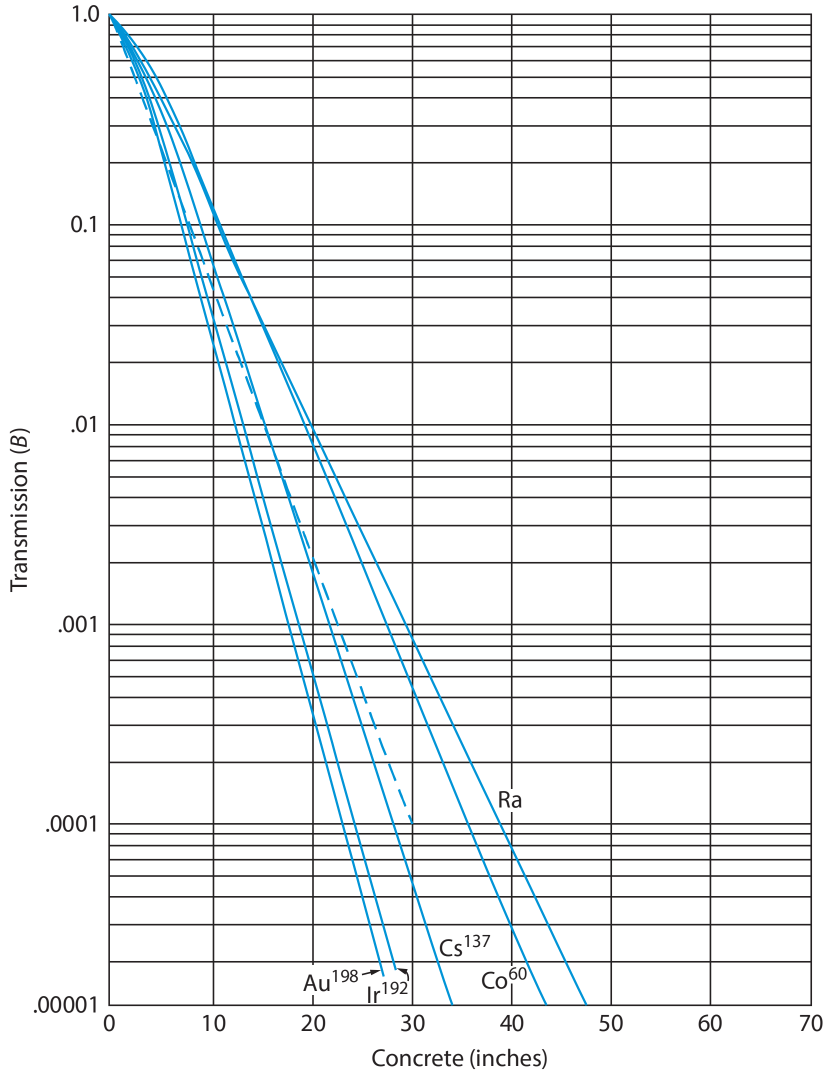

Figure 10-6. Fractional transmission of gamma rays from ^{137}Cs, ^{60}Co, and Ra (in equilibrium with its decay products) through concrete. The solid curves represent transmission of broad beams, and the broken line represents ^{60}Co gamma-ray attenuation under conditions of good geometry. (Reproduced from *Radiological Health Handbook*. Rev ed. Washington, DC: U.S. Government Printing Office; 1970.)

these values are given in Figures 10-9 and 10-10. In these curves, the shield thickness is given in units of *relaxation lengths*. One relaxation length is that thickness of shield that will attenuate a narrow beam to $1/e$ of its original intensity. One relaxation length, therefore, is given by

$$\text{Relaxation length} = \mu x = \ln \frac{I}{I_0}.$$

When $\mu x = 1$, we have 1 relaxation length, which can also be referred to as a mean free path length.

The use of the buildup factor in the calculation of a shield thickness may be illustrated with the following examples.

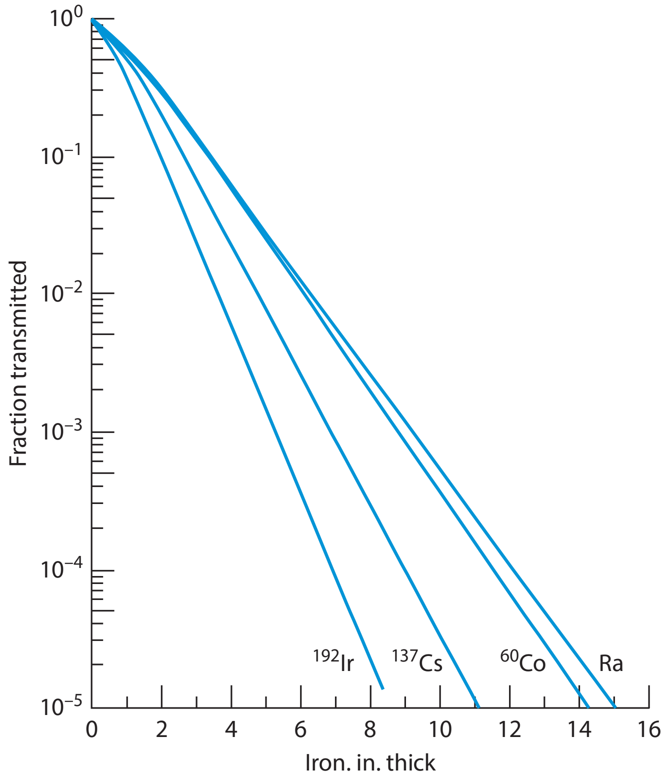

Figure 10-7. Broad-beam attenuation by iron of gamma-rays from ^{192}Ir, ^{137}Cs, ^{60}Co, and radium. (Reproduced from *Radiological Health Handbook*. Rev ed. Washington, DC: U.S. Government Printing Office; 1970.)

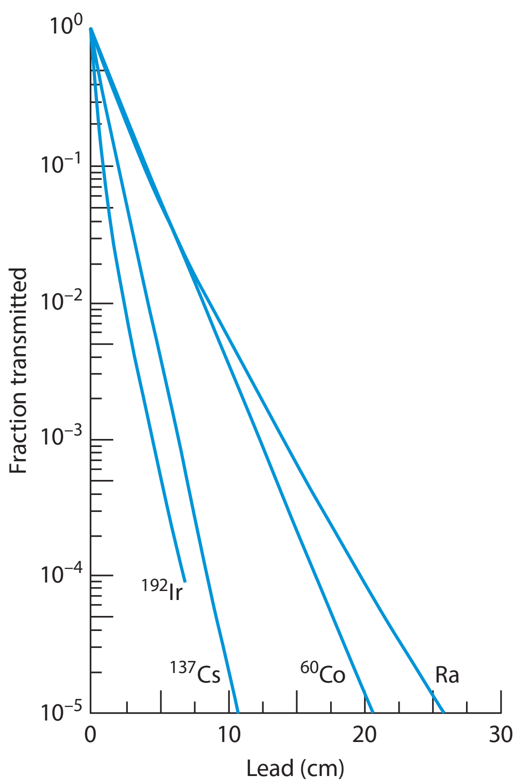

Figure 10-8. Broad-beam attenuation by lead of gamma-rays from ^{192}Ir, ^{137}Cs, ^{60}Co, and radium. (Reproduced from *Protection Against Radiations from Sealed Gamma Sources*. Washington, DC: National Bureau of Standards; 1960. NBS Handbook 73.)

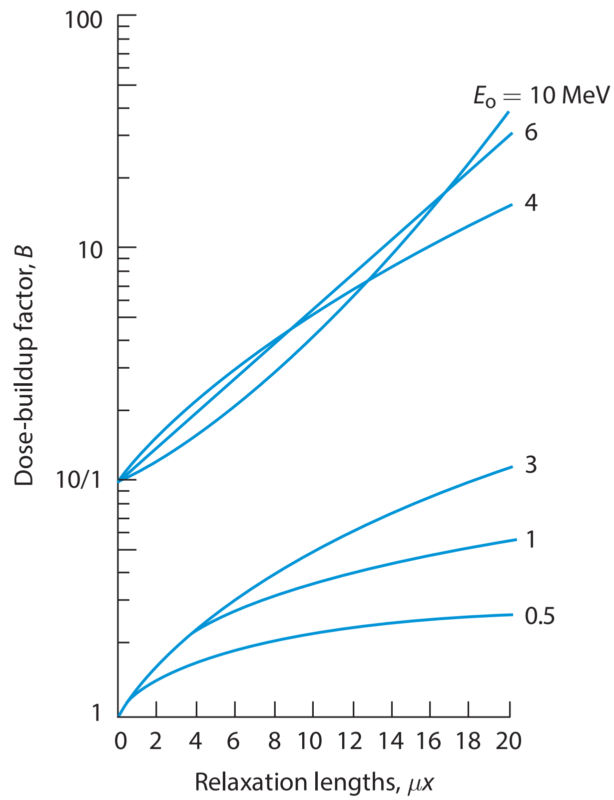

Figure 10-9. Dose-buildup factor in lead for a point isotropic gamma-ray source of energy E_0. The 10 in 10/1 on the ordinate applies to the lower curves, while the 1 in 10/1 on the ordinate applies to the upper curves. (Reproduced from *Radiological Health Handbook*. Rev ed. Washington, DC: U.S. Government Printing Office; 1970.)

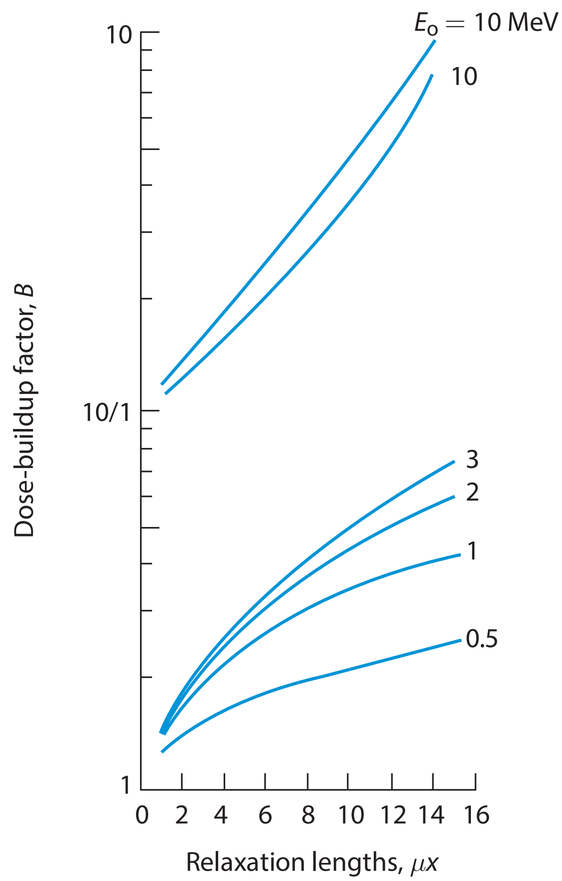

Figure 10-10. Dose-buildup factor in lead for a plane monodirectional gamma-ray source of quantum energy E_0. The 10 in 10/1 on the ordinate applies to the lower curves, while the 1 in 10/1 on the ordinate applies to the upper curves. (Reproduced from *Radiological Health Handbook*. Rev ed. Washington, DC: U.S. Government Printing Office; 1970.)

In Figures 10-9 and 10-10, we see that the buildup factor is a function of the shield thickness. Since the shield thickness is not yet known, Eq. (10.18) has two unknowns, the buildup factor *B* and the shield thickness *t*. To determine the proper shield thickness, we estimate a thickness, then substitute this estimated value into Eq. (10.18) to determine whether it will satisfy the dose-rate reduction requirement. The minimum shield thickness can be estimated by assuming narrow-beam attenuation and then increasing the thickness thus calculated by one-half value layer (1 HVL) to account for buildup. The HVL of lead for 0.661-MeV gamma rays is

$$\mathrm{HVL} = \frac{\ln 2}{\mu} = \frac{0.693}{1.42\ \mathrm{cm}^{-1}} = 0.49\ \mathrm{cm}.$$

The estimated shield thickness therefore is 3.4 plus 0.49, or 3.9 cm, which corresponds to $1.42\ \mathrm{cm}^{-1} \cdot 3.9\ \mathrm{cm} = 5.5$ relaxation lengths. From Figure 10-9, we find (by interpolation) the dose-buildup factor for 0.661-MeV gamma rays to be 2.12 for a lead shield of this thickness. Substituting these values for *B* and *x* into Eq. (10.18), we have

$$\begin{aligned} I &= 3300 \frac{\mu\mathrm{Sv}}{\mathrm{h}} \cdot 2.12 \cdot e^{-1.42\ \mathrm{cm}^{-1} \cdot 3.9\ \mathrm{cm}} \\ &= 27.5 \frac{\mu\mathrm{Sv}}{\mathrm{h}} \ \left(2.75\ \frac{\mathrm{mrems}}{\mathrm{h}}\right). \end{aligned}$$

This calculated reduction in gamma-ray dose rate is just slightly more than the desired value of 25 μSv/h (2.5 mrems/h). The thickness of 3.9 cm, as calculated above, is therefore just about correct. In this example, we found the correct thickness after one attempt in a trial-and-error method. If the calculated reduction in radiation dose rate had not turned out to be so close to the design value with the estimated shield thickness, we would have continued, by trial and error, to estimate thicknesses until the one that results in the desired reduction of dose rate was obtained.

X-Rays

In Chapter 6 we saw that the roentgen, which had been originally used as a dosimetric quantity, was really a measure of X-ray exposure rather than dose. However, the roentgen continued to be used as a dosimetric quantity because an exposure of 1 R, which deposited 87.8 ergs of energy to a gram of air, deposited 97 ergs of energy to a gram of soft tissue. Since an absorbed dose of 1 rad corresponds to absorption of 100 ergs/g, the absorbed dose from a 1-R exposure is approximately 1 rad. Furthermore, since the radiation weighting factor $w_R = 1$ for X-rays, a 1-rad X-ray dose = 1-rem dose equivalent. Thus, a 1-R exposure leads to an approximate dose equivalent of 1 rem, or, in SI units, to a dose equivalent of 1 cSv (or 10 mSv). However, the roentgen is now an obsolete unit. For X-ray protection purposes, the quantity *air kerma* is often used. This unit is especially useful to express exposure in SI units because exposure measured in air kerma is considered numerically equal to the dose equivalent measured in Sv.

Shielding for protection against X-rays is considered under two categories: source shielding and structural shielding. Source shielding is usually supplied by the manufacturer of the X-ray equipment in the form of a lead shield in which the X-ray tube is housed. The safety standards recommended by the National Council on Radiation Protection and Measurements (NCRP) specify the following types of protective tube housings for medical X-ray installations (NCRP 102):

1. *Diagnostic type*: It is so built that the leakage-radiation air kerma at a distance of 1 m from the target cannot exceed 1 mGy (100 mrads) in 1 hour when the tube is operated at its maximum continuous rated current and high voltage.
2. *Therapeutic type*:
   - **a.** For X-rays generated at voltages of 5 to 50 kV—The tube housing is built so that the maximum-leakage kerma rate at any point 5 cm from the tube housing does not exceed 1 mGy (100 mrads) in 1 hour when the tube is operated at its maximum rated beam current and high voltage.
   - **b.** For X-rays generated at voltages greater than 50 kV but less than 500 kV—A tube housing built so that the leakage kerma rate at a distance of 1 m from the target does not exceed 1 cGy (1 rad) in 1 hour. Furthermore, the leakage kerma rate at a distance of 5 cm from the tube housing does not exceed 30 cGy/h (30 rads/h).
   - **c.** For X-ray generated at peak voltages of 500 kV or more—A protective source housing built so that (i) the leakage-radiation rate in a region outside of the maximum-sized

useful beam but within a 2-m radius circular plane centered on the beam's central axis at the normal treatment distance does not exceed 0.2% of the treated tissue dose rate and (ii) except for this region, the absorbed dose rate at 1 m from the electron path between the source and the target does not exceed 0.5% of the treatment dose rate on the central axis of the beam at the normal treatment distance.

For nonmedical X-rays, a protective tube housing is one that surrounds the X-ray tube itself, or the tube and other parts of the X-ray apparatus (e.g., the transformer), and is so constructed that the leakage radiation at a distance of 1 m from the target cannot exceed 1 rem in 1 hour when the tube is operated at any of its specified ratings. Leakage radiation, as used in these specifications for tube housings, means all radiation, except the useful beam coming from the tube housing.

Structural shielding is designed to protect against the useful X-rays, leakage radiation, and scattered radiation. It encloses both the X-ray tube (with its protective tube housing) and the space in which the object being irradiated is located. Structural shielding may vary considerably in form. It may, for example, be either a lead-lined box in the case of an X-ray tube used by a radiobiologist to irradiate small organisms, or it may be the shielding around a room in which a patient is undergoing diagnostic procedures utilizing radiation sources or radiation therapy. In any case, structural shielding is designed to protect people in an occupied area outside an area of high radiation intensity. The structural shielding requirements for a given installation are determined by

1. the maximum kilovoltage at which the X-ray tube is operated,
2. the maximum milliamperes of beam current,
3. the workload (W), which is a measure, in suitable units, of the amount of use of an X-ray machine. For X-ray shielding design, workload is usually expressed in units of milliampere-minutes per week,
4. the use factor (U), which is the fraction of the workload during which the useful beam is pointed in the direction under consideration, and
5. the occupancy factor (T), which is the factor by which the workload should be adjusted to correct for the degree or type of occupancy of the area in question. When adequate occupancy data are not available, the values for T given in Table 10-1 may be used as a guide in planning shielding.

According to the International Commission on Radiological Protection (ICRP) Publication No. 26 (1977)—*General Recommendations for Protection of Radiation Workers*—and ICRP 57 (1989)—*Recommendations for Protection of Workers in Medicine and Dentistry*—the annual dose limit is 50 mSv (5000 mrems). In ICRP 60 (1990), ICRP 103 (2008), and ICRP 130 (2015), the recommended dose limit was changed to 100 mSv (10,000 mrems) over a 5-year period, with a maximum dose in any single year of 50 mSv (5000 mrems). The mean annual dose limit is thus 20 mSv (2000 mrems). These recommendations have been incorporated into the radiation safety regulations of many but not all countries. Since no harmful effects have been observed among workers whose dose was limited to 5000 mrems in 1 year, radiation safety regulations in the United States continue to be based on an annual occupational dose limit of 5000 mrems (50 mSv) and 100 mrems (1 mSv) in 1 year for the nonoccupationally exposed members of the public. At a uniform rate of exposure over 50 weeks, these maxima correspond to 100 mrems (1 mSv) per week for occupational exposure and to 2 mrems (0.02 mSv) per week for individuals who are not radiation workers.

TABLE 10-1 Suggested Occupancy Factors[a]

LOCATION	OCCUPANCY FACTOR (T)
Administrative or clerical offices; laboratories, pharmacies, and other work areas fully occupied by individuals; receptionist areas, attended waiting rooms, children's indoor play areas, adjacent X-ray rooms, film-reading areas, nurse's stations, and X-ray control rooms	1
Rooms used for patient examinations and treatments	1/2
Corridors, patient rooms, employee lounges, staff rest rooms	1/5
Corridor doors[b]	1/8
Public toilets, unattended vending areas, storage rooms, outdoor areas with seating, unattended waiting rooms, patient holding areas	1/20
Outdoor areas with only transient pedestrian or vehicular traffic, unattended parking lots, unattended vehicular drop off areas, attics, stairways, unattended elevators, janitor's closets	1/40

[a]When using a low-occupancy factor for a room immediately adjacent to an X-ray room, care *should* be taken to also consider the areas further removed from the X-ray room. These areas may have significantly higher occupancy factors than the adjacent room and may therefore be more important in shielding design despite the larger distances involved.

[b]The occupancy factor for the area just outside a corridor door can often be reasonably assumed to be lower than the occupancy factor for the corridor.

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NCRP 147 Methodology[3]

Recommendations for the design of structural shielding for medical radiation facilities that had been published in NCRP 49 have been updated. The new recommendations for shielding design for medical imaging facilities are found in NCRP 147, and, for therapeutic facilities, the revised recommendations are found in NCRP 151. For X-rays, the updated recommendations specify exposure in units of air kerma, K, rather than in roentgen units, and the minimum distance from a shielded wall to an occupied area is assumed to be 0.3 m. Additionally, the calculational methodology for imaging facilities is based on mathematical models derived from extensive measurements made during a survey made by the American Association of Physicists in Medicine (AAPM) at 14 different medical institutions involving about 2500 patients and 7 types of radiological installations. Shielding design recommendations are made for each of the different types of installations. These include:

- Radiographic installations. These are general-purpose installations that employ X-ray tubes operating at potentials of 50–150 kVp (kV peak). Radiographic installations do not have provisions for fluoroscopy. Three subcategories based on use and orientation of the X-ray tube are described:
  - Rad room (all barriers), used only for secondary barriers,
  - Rad room (chest bucky), and
  - Rad room (floor and other barriers).

[3]This chapter is intended to provide an introduction to NCRP Report No. 147 on *Structural Shielding Design for Medical X-Ray Imaging Facilities*, which contains numerous worked examples and additional information not contained herein. It is recommended that the reader obtain a copy of Report No. 147 from NCRP at http://ncrponline.org/publications for a more complete treatment of shielding for medical X-ray imaging facilities.

- The Rad room (all barriers) is composed of the sum of the other two Rad rooms. These two include beams directed at the floor, or any other directions.
- The walls at which the beam is directed form the primary barriers. For this reason, the Rad room (all barriers) data are used only for the design of secondary barriers.
- Fluoroscopy installations. Since fluoroscopic units also are used for radiographic imaging, two subcategories are considered:
  - Fluoroscopy tube (R & F room). Fluoroscopy is usually done at X-ray potentials of 60–120 kVp. Since the fluoroscopic image receptor is designed as a primary barrier, all the walls are considered secondary barriers against scattered and leakage radiation.
  - Rad tube (R & F room)
- Chest room, dedicated to chest X-rays only. The image receptor is located at a particular wall, which is the primary protective barrier. All other walls are secondary barriers.
- Mammography room. Mammography is performed at X-ray voltages of 25–35 kVp, and the image receptors of dedicated mammographic units are required to intercept the primary beam. Dedicated mammography installations, therefore, may not require any more shielding than that afforded by the structural materials of the walls of the room.
- Interventional imaging facilities. Two subcategories are considered:
  - cardiac angiography
  - peripheral angiography

Per patient workload values, which are exposure values for each of these applications that are weighted according to the distribution of operating high voltages for each procedure, are listed in the NCRP report in units of milliamperes-minutes per week (Table 10-2) and are incorporated into the recommended design methodology. The workload for a given facility is defined as the total number of milliamperes-minutes per week that the X-ray tube is in operation. The average workload per patient, which may include multiple exposures due to several different radiological modalities, is called the normalized workload, W_{norm}, and the total workload for a given installation is the product of the normalized workload and the weekly number of patients, N:

$$W = N \cdot W_{\text{norm}}.$$

TABLE 10-2 Mean per Patient kVp-Weighted Workload, W_{norm}, for Several Medical Imaging Installations

INSTALLATION	W_{norm} (mA-min wk^{-1})	K_P^1 (mGy/PATIENT)	PATIENTS PER WEEK
Rad room, all barriers	2.5		110 total for all
Rad room, chest bucky	0.6	2.3	Rad room uses
Rad room floor or other barriers	1.9	5.2	
R & F room, fluoroscopy	13		18
R & F room, radiography	1.5	5.9	23
Chest room	0.22	1.2	210
Mammography room	6.7		47
Cardiac angiography	160		19
Peripheral angiography	64		21

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To decrease the unshielded primary air kerma, $K_P(0)$, at a location at a distance d_P from the X-ray tube requires a barrier that will transmit the following fraction of the unshielded incident radiation:

$$B_P = \left(\frac{P}{T}\right)\frac{d_P^2}{K_P^1 \cdot U \cdot N}, \tag{10.20}$$

where

P = allowable air kerma rate, either 1 mGy/wk for a controlled area or 0.02 mGy/wk for an uncontrolled area,
T = occupancy factor (Table 10-1),
d_P = distance from X-ray tube to point of interest (usually 1 ft, or 0.3 m, from the barrier),
K_P^1 = unshielded primary air kerma per patient at a distance of 1 m,
U = use factor = fraction of the time that the primary beam is directed toward a given primary barrier, and
N = number of patients per week.

Using the kVp weighted values for the workload, and using three empirically determined parameters, mathematical models were fitted to the data obtained for each of the exposure modalities that had been investigated, and broad-beam transmission equations were determined for primary and secondary protective barriers. The thickness of the primary barrier is given by

$$x = \frac{1}{\alpha\gamma}\ln\left[\frac{\left(\dfrac{NTUK_P^1}{Pd_P^2}\right)^{\gamma} + \dfrac{\beta}{\alpha}}{1 + \dfrac{\beta}{\alpha}}\right], \tag{10.21}$$

where α, β, and γ are parameters that depend on the barrier material and on the operating potential of the X-ray tube, which is implicit in the imaging modality. NCRP Publication 147 lists the values for these parameters for lead, concrete, gypsum wallboard, steel, plate glass, and wood. Values for lead, concrete, and gypsum wallboard for the several modalities that had been studied are listed in Table 10-3.

For calculating the thickness of a secondary barrier, the following empirical equation is used:

$$x = \frac{1}{\alpha\gamma}\ln\left[\frac{\left(\dfrac{NTK_{sec}^1}{Pd_{sec}^2}\right)^{\gamma} + \dfrac{\beta}{\alpha}}{1 + \dfrac{\beta}{\alpha}}\right]. \tag{10.22}$$

The parameter values for secondary radiation are different from those for primary radiation because of the different X-ray energies of the secondary radiation. Table 10-4 lists some of these values.

TABLE 10-3 Values for the Parameters for Transmission of Broad-Beam Primary X-rays

WORKLOAD DISTRIBUTION[a]	LEAD			CONCRETE[b]			GYPSUM WALLBOARD		
	α (mm^{-1})	β (mm^{-1})	γ	α (mm^{-1})	β (mm^{-1})	γ	α (mm^{-1})	β (mm^{-1})	γ
Rad rooms (all barriers)	2.346	1.590×10^{1}	0.4982	3.626×10^{-2}	1.429×10^{-1}	4.932×10^{-1}	1.420×10^{-2}	5.781×10^{-2}	7.445×10^{-1}
Rad rooms (chest bucky)	2.264	1.308×10^{1}	0.5600	3.552×10^{-2}	1.177×10^{-2}	6.007×10^{-1}	1.278×10^{-2}	4.848×10^{-2}	8.609×10^{-1}
Rad rooms (floor or other barriers)	2.651	1.656×10^{1}	0.4585	3.994×10^{-2}	1.448×10^{-1}	4.231×10^{-1}	1.679×10^{-2}	6.124×10^{-2}	7.356×10^{-1}
Fluoroscopy tube (R&F room)	2.347	1.267×10^{1}	0.6149	3.616×10^{-2}	9.721×10^{-2}	5.186×10^{-1}	1.340×10^{-2}	4.283×10^{-2}	8.796×10^{-1}
Rad tube (R&F room)	2.295	1.300×10^{1}	0.5573	3.549×10^{-2}	1.164×10^{-1}	5.774×10^{-1}	1.300×10^{-2}	4.778×10^{-2}	8.485×10^{-1}
Chest room	2.283	1.074×10^{1}	0.6370	3.622×10^{-2}	1.766×10^{-2}	5.404×10^{-1}	1.286×10^{-2}	3.505×10^{-2}	9.356×10^{-1}
Mammography room	30.60	1.776×10^{2}	0.3308	2.577×10^{-2}	1.765	3.644×10^{-1}	9.148×10^{-2}	7.090×10^{-1}	3.459×10^{-1}
Cardiac angiography	2.389	1.426×10^{1}	0.5948	3.717×10^{-2}	1.087×10^{-1}	4.879×10^{-1}	1.409×10^{-2}	4.814×10^{-2}	8.419×10^{-1}
Peripheral angiography[c]	2.728	1.852×10^{1}	0.4614	4.292×10^{-2}	1.538×10^{-1}	4.236×10^{-1}	1.774×10^{-2}	3.449×10^{-2}	7.158×10^{-1}
*PET 0.511 MeV	0.1543	−0.04408	2.136	0.01539	−0.0116	2.07552	-	-	-

[a]The workload distributions are those surveyed by the AAPM.

[b]The fitting parameters for concrete assume standard-density concrete.

[c]The data in this table for peripheral angiography also apply to neuroangiography.

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*For use in PET facilities: Data from Madsen MT, et al. AAPM Task Group 108: PET and PET/CT Shielding Requirements. *Med Phys*. January 2006; 33(1): 4-15.

TABLE 10-4 Values for the Parameters for Broad-Beam Secondary Transmission of X-rays[a]

WORKLOAD DISTRIBUTION[b]	LEAD			CONCRETE[c]			GYPSUM WALLBOARD		
	α (mm^{-1})	β (mm^{-1})	γ	α (mm^{-1})	β (mm^{-1})	γ	α (mm^{-1})	β (mm^{-1})	γ
30 kVp	3.879×10^{1}	1.800×10^{2}	3.560×10^{-1}	3.174×10^{-1}	1.725	3.705×10^{-1}	1.198×10^{-1}	7.137×10^{-2}	3.703×10^{-2}
50 kVp	8.801	2.728×10^{1}	2.957×10^{-1}	9.030×10^{-2}	1.712×10^{-1}	2.324×10^{-1}	3.880×10^{-2}	8.730×10^{-2}	5.105×10^{-1}
70 kVp	5.369	2.349×10^{1}	5.883×10^{-1}	5.090×10^{-2}	1.697×10^{-1}	3.849×10^{-1}	2.300×10^{-2}	7.160×10^{-2}	7.300×10^{-1}
100 kVp	2.507	1.533×10^{1}	9.124×10^{-1}	3.950×10^{-2}	8.440×10^{-2}	5.191×10^{-1}	1.470×10^{-2}	4.000×10^{-2}	9.752×10^{-1}
125 kVp	2.233	7.888	7.295×10^{1}	3.510×10^{-2}	6.600×10^{-2}	7.832×10^{-1}	1.200×10^{-2}	2.670×10^{-2}	1.079
150 kVp	1.791	5.478	5.678×10^{-1}	3.240×10^{-2}	7.750×10^{-2}	1.566	1.040×10^{-2}	2.020×10^{-2}	1.135
Rad room (all barriers)	2.298	1.738×10^{1}	6.193×10^{-1}	3.610×10^{-2}	1.433×10^{-1}	5.600×10^{-1}	1.380×10^{-2}	5.700×10^{-2}	7.937×10^{-1}
Rad room (chest bucky)	2.256	1.380×10^{1}	8.837×10^{-1}	3.560×10^{-2}	1.079×10^{-1}	7.705	1.270×10^{-2}	4.450×10^{-2}	1.049
Rad room (floor or other barriers)	2.513	1.734×10^{1}	4.994×10^{-1}	3.920×10^{-2}	1.464×10^{-1}	4.486×10^{-1}	1.640×10^{-2}	6.080×10^{-2}	7.472×10^{-1}
Fluoroscopy tube (R&F room)	2.322	1.291×10^{1}	7.575×10^{-1}	3.630×10^{-2}	9.360×10^{-2}	5.955×10^{-1}	1.330×10^{-2}	4.100×10^{-2}	9.566×10^{-1}
Rad tube (R&F room)	2.272	1.360×10^{1}	7.184×10^{-1}	3.560×10^{-2}	1.114×10^{-1}	6.620×10^{-1}	1.290×10^{-2}	4.570×10^{-2}	9.355×10^{-1}
Chest room	2.288	9.848	1.054	3.640×10^{-2}	6.590×10^{-2}	7.543×10^{-1}	1.300×10^{-2}	2.970×10^{-2}	1.195
Mammography room	2.991×10^{1}	1.884×10^{2}	3.550×10^{-1}	2.539×10^{-2}	1.8411	3.924×10^{-1}	8.830×10^{-2}	7.526×10^{-1}	3.786×10^{-1}
Cardiac angiography	2.354	1.494×10^{1}	7.481×10^{-1}	3.710×10^{-2}	1.067×10^{-1}	5.733×10^{-1}	1.390×10^{-2}	4.640×10^{-2}	9.185×10^{-1}
Peripheral angiography[d]	2.661	1.954×10^{1}	5.094×10^{-1}	4.219×10^{-2}	1.559×10^{-1}	4.472×10^{-1}	1.747×10^{-2}	6.422×10^{-2}	7.299×10^{-1}

[a]The values of parameters are applicable to barrier thicknesses within the range of Figures 10-6, 10-7, and 10-8.

[b]The 30-kVp and mammography room data are for molybdenum-anode X-ray tubes. All other data are for tungsten-anode X-ray tubes.

[c]Standard-density concrete is assumed.

[d]The data for peripheral angiography also apply to neuroangiography.

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Some of the physical factors that determine the shielding requirements for protection from X-ray beams are shown in Figures 10-11, 10-12, and 10-13. A collimated X-ray beam of area *F* is directed at patient M (or object to be radiographed, in the case of nonmedical radiography) from the shielded X-ray tube. The beam passes through the patient and is attenuated to an acceptable level by the primary protective barrier before irradiating a person in area 1. The leakage radiation from the X-ray tube and the scattered radiations are attenuated to an acceptable level by a secondary protective barrier before reaching points outside the walls where other people may be irradiated.

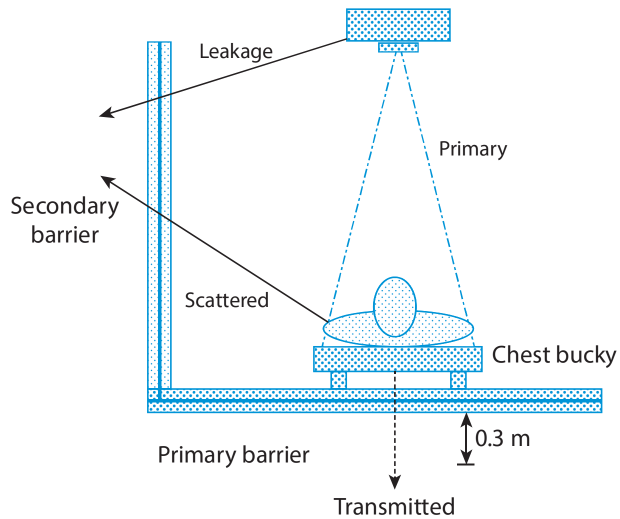

Figure 10-11. Diagram showing primary, scattered, leakage, and transmitted radiations in a radiographic room with a patient standing against the chest bucky. The distance from the shielded wall to an occupied point is assumed to be 0.3 m. (Reproduced with permission of the National Council on Radiation Protection and Measurements from *Structural Shielding Design for Medical X-Ray Imaging Facilities*. Bethesda, MD: National Council on Radiation Protection & Measurement; 2004. NCRP Report 147.)

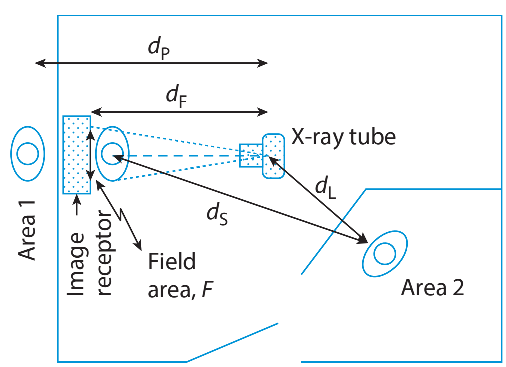

Figure 10-12. A typical medical imaging room layout. For the indicated tube orientation, the person in area 1 would need to be shielded from the primary beam, with the distance from the X-ray source to the shielded area equal to d_P. The person in area 2 would need to be shielded from scattered (distance d_S) and leakage radiations (distance d_L). The primary X-ray beam has area *F* at distance d_F. It is assumed that persons in occupied areas are at a distance of 1 ft. (0.3 m) beyond the barrier walls, 1.7 m above the floor below, and 0.5 m above occupied floor levels in rooms above the imaging room. (Reproduced with permission of the National Council on Radiation Protection and Measurements from *Structural Shielding Design for Medical X-Ray Imaging Facilities*. Bethesda, MD: National Council on Radiation Protection & Measurement; 2004. NCRP Report 147.)

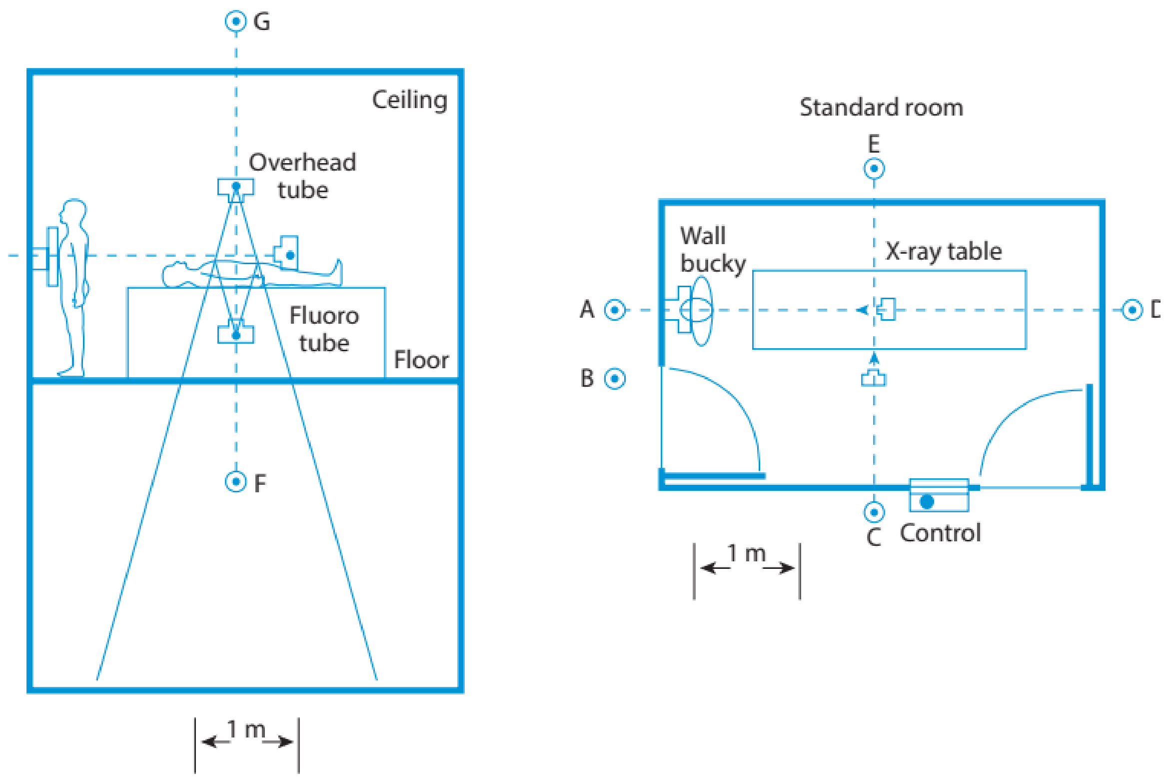

Figure 10-13. Elevation (*Left*) and plan (*Right*) views of a representative radiographic and fluoroscopic room. Points *A, B, C, D*, and *E* represent a distance of 0.3 m (1 ft) from the respective walls. Point F is 1.7 m above the floor below. Point *G* is 0.5 m above the floor of the room above. (Reproduced with permission of the National Council on Radiation Protection and Measurements from *Structural Shielding Design for Medical X-Ray Imaging Facilities*. Bethesda, MD: National Council on Radiation Protection & Measurement; 2004. NCRP Report 147.)

Computed Axial Tomography

Computed axial tomography (CAT or CT) is a diagnostic tool that uses X-rays and computers to construct a cross-sectional image of any part of the body. CAT is based on the principle that an image of an object can be constructed from attenuation of numerous X-ray beams that pass through the object from different angles. To accomplish this, CT scanners have a circular ring-shaped gantry (Figure 10-16), an X-ray source, and one or more detectors located diametrically opposite to the X-ray tube. The patient is placed inside the circular gantry, and the gantry rotates at a rate of about 60–120 RPM (1–2 revolutions per second), while the table on which the patient lies is moved horizontally through the gantry. In this mode of operation, the X-ray beam traces a helix, which allows manipulation of the resultant image.

The X-ray beam that is directed at the patient is narrowly collimated perpendicular to the axis of rotation, resulting in a narrow beam along the axis, on the order of 1–10 mm, and is fan-shaped in the radial direction. As the X-ray tube is rotated, the incident beam is

Figure 10-16. Positron emission tomography/computed axial tomography (PET CT) scanner. The CT scanner is the first "ring" and the second is the PET scanner. Colorado State University Veterinary Teaching Hospital.

attenuated in a manner dependent on the local tissue composition (greater attenuation for bones, lesser for soft tissues). These attenuation changes are measured by the detectors, and from the relationship between the signals generated by the attenuated beam and their radial distribution, a computer constructs an image representing the cross section of the scanned area. The energy of the X-ray beam (determined by tube potential and filtration) and photon fluence (determined by the product of tube current and time) are among the main factors that affect the radiation dose to the patient.

In conventional radiography, radiation dose decreases continuously from the beam's entrance into the body to its exit. In CT, the dose is distributed more uniformly across the scanning plane because the patient is equally irradiated from all directions by the rotating X-ray source. In a CT examination of the head, for instance, the dose is relatively uniform across the field of view. In larger objects such as the chest or abdomen, the dose is equally distributed around the surface of the skin, and decreases by a factor of about two near the center of the object. Dose comparisons, therefore, between CT and conventional radiography in terms of skin dose are not appropriate. Furthermore, the radiation energy delivered by CT is not fully contained within the scanning volume. Scattered radiation, divergence of the radiation beam, and limits to the efficiency of beam collimation all contribute to the radiation dose outside scan volume. In the case of the multiple scans required to image some length of a patient's body, it becomes necessary to consider the radiation dose delivered beyond the boundaries of a single scan.

The principal metric used in CT dosimetry is called the CT dose index, or $CTDI_{100}$. The $CTDI_{100}$ integrates the radiation dose along the axis for the entire scan; it includes radiation

Positron Emission Tomography

Positron emission tomography (PET) is a diagnostic technique that is useful in identifying tissues in which there is a high rate of metabolism. PET differs from other radiation-based diagnostic modalities. While other imaging modalities produce an anatomic image of a tissue or organ, PET images sites of metabolic activity. The basic principle here is that the rate of cellular utilization of sugar increases with increased metabolic rate. Fluorodeoxyglucose (FDG) is an analogue of glucose, and is metabolized like sugar.

In PET applications, FDG is labeled with the 110-minute positron emitting radioisotope ^{18}F to make ^{18}F-fluorodeoxyglucose (18FDG), and the labeled 18FDG is injected into the patient. After about 45 minutes, the 18FDG will have been absorbed and widely distributed throughout the body. At sites of high metabolic activity, such as in that part of the brain where certain neurologic processes are occurring, or in malignant tumors whose growth requires a relatively high rate of sugar use, there will be a relatively greater concentration of 18FDG and its metabolic products than in other regions of the body. In conditions where there is decreased metabolic activity, such as that occurs in the brains of Alzheimer's disease sufferers and in nonviable heart muscle, the decrease from normal metabolic activity will be seen in a PET scan. This fact is used to distinguish between densities on radiographs that are not malignant, such as fibrotic tissue or scar tissue, whose cells do not metabolize rapidly, and primary or metastatic cancers whose cells do undergo rapid metabolism.

The metabolically hyperactive site is visualized in a PET scan by measuring the intensity of the 0.51-MeV gamma photons that are produced when the positrons from the ^{18}F are annihilated on interaction with the electrons within the metabolically active cells. The annihilation photons travel in a straight line in diametrically opposite directions. The patient is inserted into a circular ring, which is lined with detectors that respond to the annihilation radiation. Each pair of diametrically opposite detectors is connected to a coincidence circuit. Only coincident counts are recorded, since only coincident counts represent the photons from the annihilation of a single positron. Electronic analysis of these coincident pulses allows the construction of an image of the metabolically active sites in that section of the body that is within the detector ring. Since this is not an anatomical image of the site, the PET image can be combined with a CT image, or an MRI image, both of which are anatomical renderings of organs, to visualize the anatomical site of the abnormal metabolic activity.

The short half-life of ^{18}F made it necessary for the early PET centers to produce their own ^{18}F in specially designed cyclotrons. Now, because of the relatively widespread use of PET scanning, regional laboratories have found it feasible to produce the radiofluorine and to distribute it to the users in the region. The 110-minute half-life makes it necessary to produce much more ^{18}F activity than will be administered to a patient.

In shielding for all the other radiation-based diagnostic modalities, we deal with relatively low quantum energies, almost always $\leq$150 kV. In PET scanning, however, we must shield against high-energy gamma radiation, 0.51 MeV, or 510 keV. For this relatively high-energy radiation, the usual wall materials such as gypsum wallboard, hollow concrete, or cinder blocks provide little shielding. For example, the HVL for 150-kVp X-rays is 0.3-mm Pb and 22-mm concrete. For 0.51-MeV gammas, the HVLs are 5-mm Pb and 98-mm concrete. Additional lead shielding therefore must always be added to the walls of the PET suite. Also, in most other modalities, we deal with a small radiation source, the X-ray target, or the relatively small scattering area where the primary beam strikes the patient, which can reasonably be approximated as a "point" for the purpose of shielding design. In the case of

PET, the activity is distributed throughout the body, and we have an extended source whose radiation is partly absorbed by the body tissues.

A typical PET suite (Figure 10-19) consists of an imaging room, a patient dosing room, and a control room. The patient is injected with 10–20 mCi (370–740 MBq) 18FDG, and then is held for between 30 to 90 minutes while the FDG is distributed within the patient's body. Considering the radioactive decay of the ^{18}F, the activity in the patient when he leaves the dosing room, A_{dr}, is

$$A_{dr} = A_0\, e^{-\frac{0.693}{110\ \text{min}}\cdot 45\ \text{min}} = 0.75 A_0.$$

The mean activity in the patient while he is in the dosing room is

$$\bar{A}_{dr} = \sqrt{A_0 \cdot 0.75 A_0} = 0.87 A_0.$$

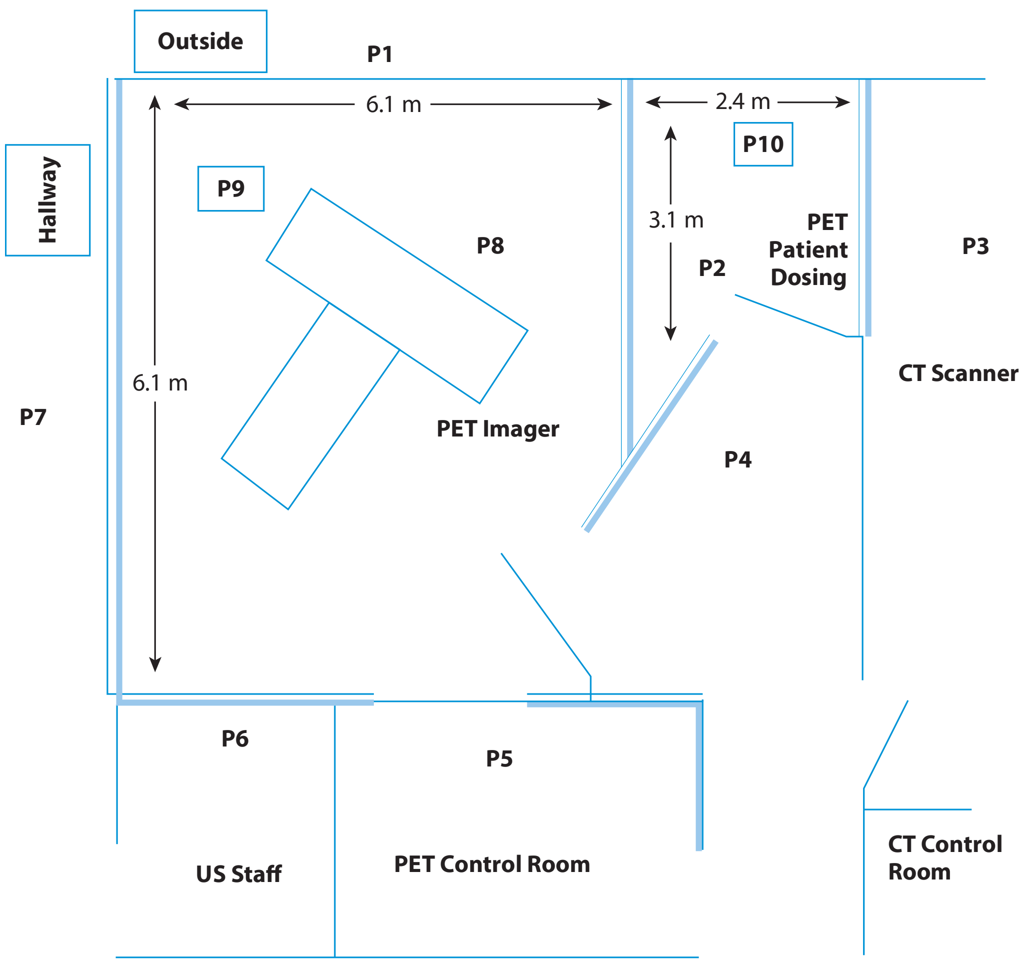

Figure 10-19. Plan view of PET suite showing physical dimensions of the imaging room and the patient dosing room. Also indicated are the dose points, P, of interest. Not drawn to scale. (Reproduced with permission from Methé BM. Shielding design for a PET imaging suite: a case study. *Health Phys*. 2003;84(5 Suppl):S83–S88.)

The reduction of activity due to the 45 minutes of decay (0.87) is called the dose reduction factor, R_t. Dose reduction factors can be calculated in a similar manner for various times for ^{18}F studies, with some listed below:

Reduction Factors, R_t, for ^{18}F

Time Post Injection (minutes)	Fraction Activity Present
30	0.91
45	0.87
60	0.83
90	0.76

If a patient receives 20 mCi (740 MBq) ^{18}F, then the average activity while in the dosing room is $0.87 \times 20 = 17.4$ mCi. About 20% of the injected FDG is excreted via the urine during the first 2 hours after injection. Thus, after 45 minutes, $45/120 \times 20\% = 7.5\%$ will have accumulated in the patient's bladder, and will have been voided before the patient enters the imaging room. This patient, therefore, will void about $0.075 \times 20 = 1.5$ mCi upon leaving the dosing room. His activity when he enters the imaging room will be $0.75 \times 20 - 1.5 = 13.5$ mCi. If 15 minutes transpire between his departure from the dosing room and the start of the scan, and if the scan lasts 45 minutes, then radioactive decay for 60 minutes will decrease the activity to $0.69 \times 13.5 = 9.3$ mCi (344 MBq), and the mean activity while the patient is in the imaging room is

$$\bar{A}_{\text{ir}} = \sqrt{13.5 \cdot 9.3} = 11.2 \text{ mCi} \quad (415 \text{ MBq}).$$

The patient's body provides some shielding for the ^{18}F 0.511-MeV annihilation radiation, and a recommended effective gamma constant that incorporates the attenuation of the body was developed by AAPM TG 108,

$$\Gamma_{\text{F-18, effective}} = 0.092 \frac{\mu\text{Sv} \cdot \text{m}^2}{\text{MBq} \cdot \text{h}}$$

$$\Gamma_{\text{F-18, effective}} = 0.34 \frac{\text{rem} \cdot \text{m}^2}{\text{Ci} \cdot \text{h}}.$$

Radiotherapy Machines[6]

Radiation from radiotherapy machines span a very wide range of energies, from 15 keV Grenz ray X-rays for superficial therapy to 25-MeV accelerators for deep therapy. The basic principle of shielding radiotherapy devices is the same as for diagnostic devices, as is the basic calculational methodology. In all cases, we

1. must know the maximum radiation level to be produced by the device,
2. calculate the unshielded radiation level at the dose point that we wish to protect,
3. calculate the required degree of attenuation, or the maximum transmission, of the shielding barrier, taking into account the use of protected area, and
4. calculate the thickness of the barrier that will give the required degree of attenuation of the radiation.

Although the basic principles are the same for all shielding-design calculations, there are significant differences in the details. Some of these differences include

1. Photoneutrons are produced when high-energy X-rays, $E > 10$ MeV, interact with matter, such as collimaters, shielding, etc. Thus, when discussing shielding of high-energy machines, we have two categories:
   a. $\leq$ 10 MeV, where we do not have neutrons because the threshold energy for photoneutron production is about 8.5 MeV for most materials, and the cross section remains very small until the quantum energy exceeds 10 MeV.
   b. $>$10 MeV, where we have neutrons that contribute significantly to the radiation dose, and which, therefore, must be shielded.
2. Because of the presence of photoneutrons, safety criteria are expressed in Sv rather than Gy.
3. Specification of the workload—For diagnostic X-ray machines, the workload is specified in milliamp-minutes per week at a given kVp. For therapeutic facilities, the workload is specified as the weekly dose at the gantry isocenter, usually at a distance of 1 m from the X-ray source, in Gy per week.
4. An additional safety requirement that the dose equivalent in any unrestricted area be $\leq$0.02 mSv (2 mrems) in any 1-hour period and a weekly requirement equal to 1/50 of the design goal. This additional requirement is called the *time averaged dose-equivalent rate* (TADR), and is 1-hour dose averaged over a period of 1 week (40 hours). The reason for the averaging is that instantaneous measurements will yield unrealistic average doses,

[5]Methe, Brian M. *Oper Rad Saf*, S83-S88, May, 2003.

[6]This introduction to radiotherapy machine shielding design is based on NCRP Report 151, *Structural Shielding Design and Evaluation for Megavoltage X- and Gamma Ray Radio Therapy Facilities*. Complete detailed information can be found in NCRP 151.

and unreasonably large shields. The weekly TADR, R_W, for a dose point behind a primary barrier is given by

$$R_W = \frac{\mathrm{IDR}\,\frac{\mathrm{Sv}}{\mathrm{h}}\cdot W_{\mathrm{pri}}\,\frac{\mathrm{Gy}}{\mathrm{wk}}\cdot U_{\mathrm{pri}}}{\dot{D}_O\,\frac{\mathrm{Gy}}{\mathrm{h}}}, \tag{10.31}$$

where

$\dot{D}_0$ = maximum absorbed dose output rate at 1 m, Gy/h,
W_{pri} = primary barrier weekly workload, Gy/wk,
U_{pri} = use factor for that dose point, and
IDR = instantaneous dose-equivalent rate, Sv/h, at 30 cm beyond the shield, operating at $\dot{D}_0$. Accelerator measurements are averaged over 20 to 60 seconds, based on duty cycle of the accelerator and the instrument ability to respond.

A generic layout of a typical high-energy treatment room is shown in Figure 10-20.

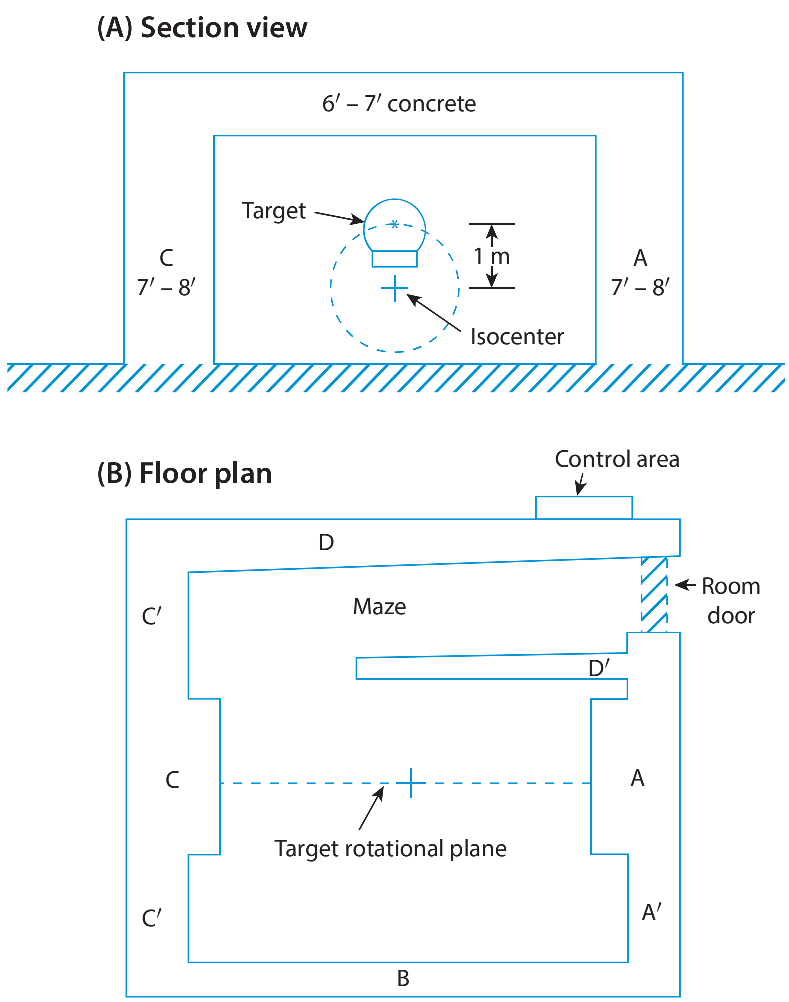

Figure 10-20. Simplified schematic of a typical high-energy treatment room. All barriers are constructed of standard concrete (147 lb ft^{-3}). (Reproduced with permission from McGinley PH. *Shielding Techniques for Radiation Oncology Facilities*. 2nd ed. Madison, WI: Medical Physics Publishing Corp; 2002: Fig 2-1, p. 10.)