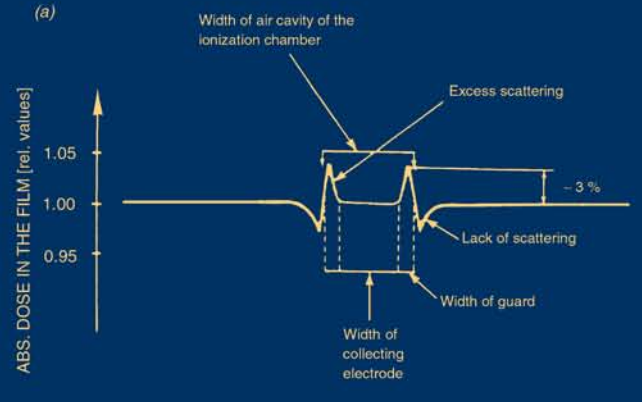
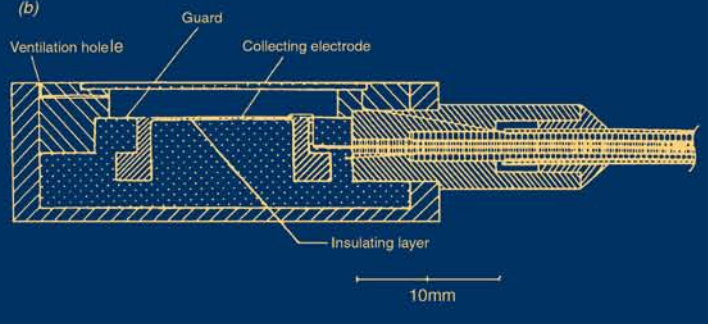
RADIATION DOSIMETRY

Instrumentation and Methods

Second Edition





Gad Shani

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Preface

This volume is an updated reference book for medical dosimetry. It evolved from the book *Radiation Dosimetry Instrumentation and Methods* (CRC Press, 1991) published 10 years ago, which contains many of the basic facts of radiation dosimetry techniques. The present book contains developments in the last decade, mainly for medical dosimetry. The two books are complementary.

Radiation dosimetry has made great progress in the last decade, mainly because radiation therapy is more widely used. Every medium to large sized hospital has an oncology department with at least one, generally more, linear accelerators for tumor treatment. Radiation dosimetry has become a common need in the medical world. Medical physicist is now a certified profession and is required by the law in every hospital where radiation treatment is given to patients. One of the main tasks of the medical physicist is to provide the physician with an accurate measurement of the dose delivered to the patient. Several measurement methods were developed together with improved calculation methods, correction factors, and Monte Carlo simulation.

It was the intention of the author to assemble this past decade's developments in one short volume. Unfortunately, because of the vast amount of material, only selected information could be included, and many important developments in this field had to be left out.

The book starts with a short introductory chapter where the basic concepts of radiation and dosimetry are defined. The second chapter deals with basic concepts of radiation dosimetry theory. Electron and photon beams are defined and their interaction with the detector material is described with the consequences of that interaction for dosimetry. Theoretical methods for calculating these effects are discussed.

The main tool in medical dosimetry is the ionization chamber. Because of the importance of an accurate dose measurement, several types of ionization chambers were developed for this purpose in the last decade, having different shapes and dimensions and made of various materials. Application of a dosimeter in a measured system causes perturbation; also the dosimeter itself is not an ideal system. The difference between the measured dose and the calculated one requires the use of correction factors. These are calculated using models developed in the last few years. Ion chambers are also used in portable monitors for area survey; some examples are discussed in Chapter 3.

Chapter 4 deals with the new developments in thermoluminescent dosimetry (TLD). The basic concepts were discussed in *Radiation Dosimetry Instrumentation and Methods* and are not repeated here. New developments of TLD in the last decade are shown with scientific results and applications. Other luminescence dosimetry methods and electron spin resonance dosimetry are included in this chapter.

The new development in film dosimetry is mainly radiochromic film. This new type of film has a high spatial resolution and low spectral sensitivity; therefore, it is useful for dose distribution measurement. Again, the basic aspects of film dosimetry are not repeated in Chapter 5.

Some progress in calorimetry dosimetry was made. However, the two materials used for dose measurement are still water and graphite, because of the similarity between these materials' interaction with radiation to that of tissue. Calorimetry dosimetry and chemical dosimetry are discussed in Chapters 6 and 7, respectively.

Solid-state dosimetry has made considerable progress in the last decade. First, the use of diamonds became popular, mainly because diamonds are made of pure carbon and are close to tissue equivalent. Also as a perfect crystal, a diamond makes a good solid-state detector. The development of solid-state devices for electronics found its way to radiation dosimetry too. MOSFET and other devices are small and capable of detecting very low currents when used as solid-state detectors. Another way of using them: radiation effect on the device's performance is an indication of the dose absorbed.

Chapter 9 deals with a new development in radiation dosimetry, a three dimensional dosimeter. Detection of changes taking place in gel is a measure of the dose absorbed. One way to detect changes is by introducing ferrous sulfate to the gel and measuring light absorption following the irradiation, as done with the Fricke dosimeter. Another way is by changing the gel properties with radiation, then using three dimensional nuclear magnetic resonance (NMR) to measure the changes. It seems to give a good solution to the problem of three dimensional dosimetry, done in one measurement. This new dosimetry method is at its early stages of application.

Chapter 10 deals with neutron dosimetry. As is well known, neutron detection cannot be done by direct ionization; as in other kinds of radiation, secondary radiation is used. Neutron measurement is done using all the techniques used for other radiation measurements, i.e., ionization chamber, scintillator (TLD for dosimetry) and solid-state detectors. In addition, methods specific to neutron measurement include activation analysis and track detector. Because neutron

interactions with matter are different from interactions with other forms of radiation, it is important for neutron dosimetry to use detectors made of tissue equivalent materials. When gamma radiation is present, pair detectors are used, and the gamma dose is subtracted. When TLD is used, it should contain isotopes with high cross-section for neutron interaction, such as Li^6 for thermal neutrons.

A few new developments in neutron dosimetry came about in the last decade. One of them is the superheated drop detector in which the number of bubbles formed in the liquid is a function of neutron dose. Solid state diodes were developed with polyethylene or boron converters. Gamma dose is subtracted in these detectors by the pulse shape discrimination technique.

Bonner sphere was developed many years ago as an area monitor. In the last decade, better designs were made with better energy response.

The book was written for medical physicists, engineers, and advanced dosimetrists. As mentioned above, it contains only the last decade's developments. If more basic data or methods are sought, the reader should refer to the book *Radiation Dosimetry Instrumentation and Methods.*

The Author

Gad Shani, Ph.D., is a professor in the department of nuclear engineering and the head of the biomedical engineering program at Ben Gurion University, Beer Sheva, Israel. Professor Shani received his B.Sc. degree in electrical engineering and M.Sc. degree in nuclear science at the Technion Israel Institute of Technology in 1964 and 1966, respectively, and his Ph.D. in nuclear engineering from Cornell University, Ithaca, New York. On the faculty at Ben Gurion University since 1970, Dr. Shani was one of the founders of the department of nuclear engineering and the head of the department from 1980 to 1984. Since 1994, he has served as the head of the biomedical engineering program.

Professor Shani is the incumbent of the Davide and Irene Sala Chair in Nuclear Engineering and the head of the Center for Application of Radiation in Medicine. He is a member of the International Euroasian Academy of Science. Professor Shani was a visiting scientist in many laboratories around the world, including the KFA in Germany as a winner of the European Community grant; in Harwell, England as a winner of the British Royal Society grant; at McMaster University in Canada; and at UCLA in the U.S. He spent sabbatical years at the University of California, Santa Barbara, Ohio State University, and Brookhaven National Laboratory (BNL) where he was a collaborator for the last 10 years. Additionally, he spent a few months every year at BNL doing research in the fields of neutron capture therapy, synchrotron radiation, and other methods for cancer treatment with radiation.

Professor Shani's field of scientific activity over the years spans a wide area, including neutron physics, reactor physics, nuclear instrumentation, dosimetry, and medical physics. His professional affiliations include the American Nuclear Society, Israel Nuclear Society (council member), Israel Ecology Society, Israel Society for Medical and Biomedical Engineering (council member), International Federation for Medical and Biomedical Engineering, Israel Society of Medical Physics, Israel Society of Radiation Research, and International Society for Neutron Capture Therapy. He has published about 150 papers in journals and meeting proceedings, authored 4 books, and co-authored 2 books.

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1 Introduction

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I. UNITS AND DEFINITIONS

The energy imparted by ionizing radiation to matter of a given mass is the fundamental quantity of radiation dosimetry. Radiation field can be described by the average number of rays (or particles) per unit area, per unit time at each point. The rays can be in a parallel beam at angle θ to the plane or moving at all directions. In the second case the examined surface should be a sphere. If the examined area is a circle with area Δa (or one quarter of the sphere area) and the number of rays crossing it at time Δt is ΔN, then the flux density is given by

$$\Phi = \frac{\Delta N}{\Delta a \Delta t} \tag{1.1}$$

and integration over time gives the fluence

$$\Phi = \frac{\Delta N}{\Delta a} \tag{1.2}$$

Another way to deal with a radiation field is by summing the kinetic energy of all the particles entering the sphere:

$$\Delta E = \sum_i T_i \tag{1.3}$$

where T_i is the kinetic energy of the ith ray or particle. The intensity is given by

$$I = \frac{\Delta E}{\Delta a \Delta t} \tag{1.4}$$

If there is more than one kind of ray or particle, the flux density and the energy fluence can be calculated for each separately. If the particles (or rays) have different energies within a range, the flux density will be the integration of the distribution (or spectrum) over the energy range. When a beam of radiation encounters matter, it will be attenuated by the interaction with the matter. The attenuation can be measured by the reduction in number of rays or panicles, or by the reduction of the total beam energy.

The official units used in health physics and dosimetry are those agreed upon by the International Commission

on Radiological Units and Measurements (ICRU). [1] In radiation protection the term for the linear energy transfer dependent factor is the quality factor (QF) by which absorbed doses are multiplied to obtain a quantity expression of the irradiation incurred on a common scale. The distribution factor (DF) expresses the nonuniformity effect of the irradiation. The product of the absorbed dose (D) and the two factors above is the dose equivalent

$$\mathrm{DE} = \mathrm{D} \times \mathrm{QF} \times \mathrm{DF} \quad (\mathrm{Sv}) \tag{1.5}$$

The units generally used in dosimetry are gray (Gy) for absorbed dose, roentgen (R) for exposure, and curie (Ci) for activity. Definitions of some terms used in dosimetry are listed below:

- Direct ionizing particles—charged particles having sufficient kinetic energy to produce ionization
- Indirect ionizing particles—uncharged particles that can produce ionizing particles
- Ionizing radiation—radiation consisting of directly and indirectly ionizing particles
- Energy imparted by ionizing radiation—the difference between the sum of energies of ionizing particles entering a certain volume and the sum of energies leaving the volume, less the energy spent in increasing any rest mass
- Absorbed dose—the quotient of the energy imparted by ionizing radiation and the mass of this volume:

$$D = \frac{\Delta E}{\Delta m} \tag{1.6}$$

  The units of absorbed dose are 1 Gy = 100 rad where 1 rad = 100 erg/gm.
- Absorbed dose rate—the quotient of the incremental absorbed dose and the absorption time:

$$\dot{D} = \frac{\Delta D}{\Delta t} \tag{1.7}$$

  The units can be Gy/min, Gy/sec, etc.
- Particle fluence—the quotient of the number of panicles ΔN that enter a sphere of area $4\Delta a$ (a test sphere) and the area Δa (the sphere cross-section area):

$$\Phi = \frac{\Delta N}{\Delta a} \tag{1.8}$$

- Particle flux rate—the incremental particle flux per time interval; ϕ denotes flux distribution with respect to energy, direction, etc.:

$$\Phi = \frac{\Delta \phi}{\Delta t} \tag{1.9}$$

- Energy fluence—the incremental kinetic energy of all particles entering the sphere of area $4\Delta a$ (cross-section Δa) per cross-section area:

$$F = \frac{\Delta E}{\Delta a} \tag{1.10}$$

- Energy flux density—the incremental energy fluence per time interval:

$$I = \frac{\Delta F}{\Delta t} \tag{1.11}$$

- Kerma—the incremental kinetic energy of all charged particles liberated by ionizing particles in a volume element divided by the mass of this volume element:

$$K = \frac{\Delta E}{\Delta m} \tag{1.12}$$

- Kerma rate—incremental kerma in time interval Δt:

$$\frac{\Delta K}{\Delta t} \tag{1.13}$$

- Exposure—the ratio between the sum of secondary electrical charge (ions of one sign produced when electrons produced by photons are stopped) in a volume element of air to the mass of that volume:

$$X = \frac{\Delta Q}{\Delta m} \tag{1.14}$$

  The unit of exposure is roentgen: 1 R = 2.58×10^{-4} Cb/kg. (This is identical to 1 ESU per 1 cc [0.001293 g] of air.)
- Exposure rate—the incremental exposure in time interval Δt:

$$\frac{\Delta X}{\Delta t} \tag{1.15}$$

  The units are R/sec (or R/min, etc.)
- Mass attenuation coefficient—the property of the material defined by

$$\frac{\mu}{\rho} = \frac{1}{\rho N}\frac{dN}{dl} \tag{1.16}$$

for indirectly ionizing particles; ρ is the material density, N is the number of particles incident normal to the material, and dN is the number of particles interacting in thickness dl.

- Mass energy transfer coefficient—the property of the material defined by

$$\frac{\mu_k}{\rho} = \frac{1}{\rho E}\frac{dE}{dl} \tag{1.17}$$

where E is the sum of kinetic energies T_i of indirectly ionizing particles meeting normally on the material of density ρ. dE is the sum of the kinetic energies of all the charged particles liberated in thickness dl. One use of this quantity is the ratio between fluence and kerma:

$$F = K\frac{\mu_k}{\rho} \tag{1.18}$$

- Mass absorption coefficient—the property of the material defined as

$$\mu_{en} = \frac{\mu_k}{(1-g)} \tag{1.19}$$

where g is the part of the energy of the secondary charged particles lost by *bremsstrahlung*.

- Mass stopping power—the property of the material defined as

$$S = \frac{dE_s}{dl} \tag{1.20}$$

where dE_s is the average energy lost by a charged particle traversing the length dl.

- Linear energy transfer—the energy imparted from charged particles to the medium

$$L = \frac{dE_L}{dl} \tag{1.21}$$

where dE_L is the average local energy imparted when the particle travels a distance dl.

- Average energy expended in a gas per ion pair formed—

$$W = \frac{E}{N_w} \tag{1.22}$$

where E is the particle initial energy and N_w is the average number of ion pairs formed by complete stopping of the particle. Activity units are

$$1\ \text{Ci} = 3.7 \times 10^{10}\ \text{sec}^{-1}$$

II. ABSORBED DOSE IN TERMS OF EXPOSURE AND STOPPING POWER

When the exposure is 1 R, the energy absorbed in air is 87.7 erg/g. The absorbed dose is

$$D_{air} = 0.877\ R\ \text{rad} \tag{1.23}$$

where R is the number of roentgens. If the medium is not air, then

$$D_m = 0.877\ R\ \frac{\mu_{en}/\rho_m}{\mu_{en}/\rho_{air}} = fR\ \text{rad} \tag{1.24}$$

where f is the number of rad per roentgen in the medium.

When the spectrum is continuous, integration should be carried out:

$$D = \int_0^\infty R(E) f(E)\, dE \tag{1.25}$$

and

$$R = \int_0^{E_{max}} R(E)\, dE \tag{1.26}$$

The absorbed dose for a charged particle can be expressed in terms of the stopping power. If the stopping power is

$$S(T) = \frac{dT}{dx} \tag{1.27}$$

and the particle fluence is $\phi(E)$, then using the definition,

$$D = \frac{\Delta E}{\Delta m} \tag{1.28}$$

$$D = \frac{1}{\rho}\int_0^{T_{max}} S(T)\phi(T)\, dt \tag{1.29}$$

where ρ is the stopping material density and the charged particles impinge perpendicular to the area.

The mean absorbed dose, D_T, in a specified tissue or organ, T, is given by

$$D_T = \frac{1}{m_T}\int_{m_T} D\, dm \tag{1.30}$$

where m_T is the mass of the tissue or organ and D is the absorbed dose in the mass element dm. The mean absorbed dose, D_T, in a specified tissue or organ equals the ratio of

the energy imparted, ε_T, to the tissue or organ, and m_T, the mass of the tissue or organ.

III. LINEAR ENERGY TRANSFER

Linear energy transfer (LET) denotes the energy lost by a charged particle per unit distance of medium traversed:

$$L = \frac{dE_L}{dl} \tag{1.31}$$

where dE_L is the average energy locally imparted to the medium. When a nonmonoenergetic radiation interacts with material, there is a distribution of LET. If the distribution of tracks is $T(L)$, then the average LET can be defined as:

$$\overline{L_T} = \int_0^{L_{max}} T(L) L \, dL \tag{1.32}$$

or dose average

$$\overline{L_D} = \int_{L_{min}}^{L_{max}} D(L) L \, dL \tag{1.33}$$

Charged particles lose energy by colliding with the atomic electrons and transferring energy to them. This energy can be half the partial energy if the particle is an electron and four times the relative mass between the electron and the particle for heavy particles. The scattered electrons that are δ-rays form their own track, which might branch to a ternary track. δ-rays of energy above 100 eV are generally considered separate particles (in some cases higher energy is taken). The selection of the lower limit of δ-rays affects the LET of the original particle and makes the calculation complicated.

Energy transfer of heavy charged particles (HCP) to nm-size targets have been investigated by Iwanami and Oda [2], taking into account δ-ray generation by HCP as well as associated δ-rays. The energy transfer into the target is mainly due to ionizing collisions of HCP with matter. Secondary electrons generated by ionizing collisions within the target, whose ranges are much larger than the target size, deposit almost all of their energy outside the target. The ionizing collisions generating such secondary electrons are therefore excluded from the energy transferred into the target and are regarded as generating a new electron fluence. The energy of these electrons is greater than the cutoff energy for δ-rays, Δ. Secondary electrons with energy less than Δ dissipate their energy locally at their production site. ICRU [3] defined two kinds of LET: unrestricted and restricted. The unrestricted LET, L_∞, is the quotient of dE and dl, where dE is the mean energy lost by a charged particle due to collisions with electrons in traversing a distance dl; thus,

$$L_\infty = \frac{dE}{dl} \tag{1.34}$$

L_∞ does not take into account δ-ray production. The restricted LET, L_Δ, is the quotient of dE by dl, where dE is the energy lost by a charged particle in traversing a distance dl due to those collisions with electrons in which the energy loss is less than the restricted energy Δ:

$$L_\Delta = \left(\frac{dE}{dl}\right)_\Delta \tag{1.35}$$

Where Δ is the cut-off energy for δ-rays and restricted energy of L_Δ,

A simplified parameter, the event size Y, was suggested by Rossi. [4] It is the ratio between the energy deposited in a small sphere by the primary and secondary particles to the sphere diameter d:

$$Y = \frac{E}{d} \tag{1.36}$$

The complication here is that Y requires additional parameter d.

A distribution of event size Y can be found as a distribution of the LET or track length. The relation between the distribution of the absorbed dose in Y, $D(Y)$, and the distribution in L, $D(L)$, can be found by examining the relation of a track length within a sphere and the sphere diameter d:

$$D(Y) = \frac{3Y^2}{L^2} \tag{1.37}$$

where Y is idealized by assuming that the tracks are straight lines, the energy loss is uniform, and Y is independent of d. $Y \leq L$ since Y_{max} occurs along the diameter, at which position $Y = L$. It is also possible to write

$$D(Y) = 3Y^2 \int_Y^{L_{max}} \frac{D(L)}{L^3} \, dL \tag{1.38}$$

A. Dose-Equivalent Quantities

A quality factor, Q, is introduced to weight the absorbed dose for the biological effectiveness of the charged particles producing the absorbed dose. It is formulated to take account of the relative effectiveness of the different types of ionizing radiation at the low exposure levels encountered in routine radiation protection practice. The quality

factor, Q, at a point in tissue is given by

$$Q = \frac{1}{D}\int_L Q(L)D_L\,dL \tag{1.39}$$

where D is the absorbed dose at that point, D_L is the distribution of D in linear energy transfer L, and $Q(L)$ is the corresponding quality factor at the point of interest. The integration is to be performed over the distribution D_L, due to all charged particles, excluding their secondary electrons.

B. Dose Equivalent

The dose equivalent, H, is the product of Q and D at a point in tissue, where D is the absorbed dose and Q is the quality factor at that point; thus,

$$H = QD \qquad \text{(Sv)} \tag{1.40}$$

The quantity dose equivalent is defined for routine radiation-protection applications. The dose equivalent, H, at a point is given by

$$H = \int_L Q(L)D_L\,dL \tag{1.41}$$

where $Q(L)$ is the quality factor for particles with linear energy transfer L and D_L is the spectral distribution, in terms of L, of the absorbed dose at the point.

C. Ambient Dose Equivalent

The ambient dose equivalent, $H^*(d)$, at a point in a radiation field is the dose equivalent that would be produced by the corresponding expanded and aligned field in the ICRU sphere [5] at a depth d on the radius opposing the direction of the aligned field. [6]

For strongly penetrating radiation, a depth of 10 mm is currently recommended. The ambient dose equivalent for this depth is then denoted by $H^*(10)$. For weakly penetrating radiation, depths of 0.07 mm for the skin and 3 mm for the eye are employed, with analogous notation.

Measurement of $H^*(10)$ generally requires that the radiation field be uniform over the dimensions of the instrument and that the instrument have an isotropic response.

D. Directional Dose Equivalent

The directional dose equivalent, $H'(d, \Omega)$ (Sv), at a point in a radiation field is the dose equivalent that would be produced by the corresponding expanded field in the ICRU sphere at a depth d on a radius in a specified direction, Ω. The ICRU sphere is a 30-cm-diameter tissue-equivalent sphere with a density of 1 g cm^{-3} and a mass composition of 76.2% oxygen, 11.1% carbon, 10.1% hydrogen and 2.6% nitrogen.

IV. DOSIMETRY METHODS

A. Ionization Method

The most widely used method of dosimetry is based on ionization. The number of ion pairs produced is

$$I = \sum_i \int_B^\infty \frac{\varepsilon_i n_i(E)\,d\varepsilon_i}{w(\varepsilon_i)} \tag{1.42}$$

where B is the lower limit of energy loss and $w_i\,(\varepsilon_i)$ is the energy required for a particle of type i at energy ε to produce an ion pair. Since for many gases w is independent of i and ε, $I = \varepsilon T/w$.

When measurement of the dose at a specific position is required, the detector dimensions must be small compared to the attenuation length of the primary radiation. If this is impossible, the first collision dose in the detector must be the same as in the medium, or, at least, the ratio between the first collision doses in the two materials must be independent of energy. It is always required that the ratio of the stopping powers in the two materials is independent of energy.

B. Chemical Methods

In some systems the chemical composition is changed by the absorbed radiation (including photographic film). If Y is the observed chemical change, then

$$Y = \sum_i \int_0^\infty \varepsilon_i n_i(\varepsilon_i, E) G_i(\varepsilon_i)\,d\varepsilon_i \tag{1.43}$$

where $G_i\,(\varepsilon_i)$ is the yield per unit energy absorbed. If G is independent of particle type and ε_i, then

$$Y = G\varepsilon T \tag{1.44}$$

C. Calorimetric Methods

The radiation energy absorbed in the dosimeter changes into thermal energy and raises the dosimetry temperature. The temperature change is given by

$$\Delta T = \frac{1}{c}\sum_i \int_0^\infty \varepsilon_i n_i(\varepsilon_i, E) F_i(\varepsilon_i)\,d\varepsilon_i \tag{1.45}$$

where $\varepsilon_i n_i\,(\varepsilon_i, E)\,d\varepsilon_i$ is the amount of energy absorbed in a unit mass. $F_i(\varepsilon_i)$ is the fraction of charged particle energy

that is degraded to heat. c is the thermal capacity of the substance. $F_i(\varepsilon_i)$ is approximately constant near unity so that

$$\Delta T = \frac{\varepsilon T}{c} \tag{1.46}$$

D. Thermoluminescence Methods

When radiation is absorbed by an impure crystal, some of the electrons are trapped in the levels created within the forbidden gap. When those electrons are forced by heat to return to the valence band, their energy is emitted as light. The total amount of light emitted is proportional to the dose absorbed in the crystal:

$$L = \sum_i \int_B^\infty \varepsilon_i n_i(E)\, d\varepsilon_i \tag{1.47}$$

where L is the total amount of light, ε_i is the light photon energy, and n_i is the number of light photons. B is the lower limit of light detection.

V. Gamma Dosimetry

A. Point Source Dose

If we define dose rate as the energy absorbed per unit volume per unit time, it is given that

$$D' = \mu I(E, r) \tag{1.48}$$

where $I(E, r)$ is the flux density of energy E at a distance r from a point source. If the point source strength is S, then

$$I(E, r) = \frac{S}{4\pi r^2} \tag{1.49}$$

when no attenuation in the surrounding material is assumed. With attenuation the flux is

$$I(E, r) = \frac{S}{4\pi r^2} e^{-\mu r} \tag{1.50}$$

For the dose rate to be in units of energy absorbed per unit time as defined above, the source must be expressed in units of energy

$$S = cE \text{ MeV/s} \tag{1.51}$$

where c is the source intensity in disintegration per second and E is in MeV. If the source strength c is expressed in curie, then

$$S = 3.7 \times 10^{10}\, cE \tag{1.52}$$

and the dose rate is

$$D' = 2.96 \times 10^9 \mu cE \frac{e^{-\mu r}}{r^2} \frac{\text{MeV}}{\text{cm}^2/\text{s}} \tag{1.53}$$

The total dose is obtained by time integration of the dose rate:

$$D = \int D'\, dt \tag{1.54}$$

or, if the dose rate is constant,

$$D = D't \tag{1.55}$$

Radioactive isotopes are an exponentially decaying source, so integration must be carried out for at least short-lived isotopes. When the source is other than a point source, the flux must be calculated accordingly. Self absorption should sometimes be included.

For high-dose measurement the following dosimeters are used: calorimeters, alanine/electron spin resonance (ESR) systems, liquid solutions (Fricke, ceric-cerous, dichromate), and polymer systems (polymethyl methacrylate, cellulose triacetate, radiochromic films and optical waveguides).

B. First Collision Dose

When a beam of ionizing radiation meets with a small mass so that the attenuation is small, the dose is referred to as first collision dose. It is expressed in terms of the energy imparted to a unit mass of the material per unit time per unit flux at the incident beam. An expression for the first collision dose for gamma rays of energy E is given by

$$D(E) = 1.602 \times 10^{-8} \sum_i N_i \{\tau_i(E)\varepsilon_{pe}(E) + \sigma_i(E)\varepsilon_c + \kappa_i(E)\varepsilon_{pp}\} \tag{1.56}$$

where $D(E)$ is in rad/(photon/cm^2); N_i is the number of atoms of the ith element per gram of material; $\tau_i(E)$, $\sigma_i(E)$, and $\kappa_i(E)$ are the photoelectric, compton, and pair production cross sections, respectively, in cm^2/atom of the ith element; and ε is the average kinetic energy transferred to the electron (or positron) in the effects taking place (pe is photoelectric, c is compton, and pp is pair production). In the photoelectric effect, $\varepsilon_{pe} = E - E_B$, where E_B is the electron binding energy. The kinetic energy transferred to the electron in the compton effect is

$$E_c = \frac{aE(1 - \cos\phi)}{1 + \alpha(1 - \cos\phi)} \tag{1.57}$$

where $\alpha = E/0.511$ MeV and ϕ is the Compton scattered photon angle.

For the pair production $\varepsilon_{pp} = \mathrm{E} - 1.022$. The summation in Equation (1.56) is over all elements in the absorbing material. The factor 1.602×10^{-8} converts MeV/gr to rad. At low energy the main effect is the photoelectric effect. The cross section is decreased when the energy is increased, and it has the Z^5 dependence. At energies above 0.2 MeV, the main interaction is the Compton effect and then the pair production. Both cross sections are relative to the number of electrons per unit volume; hence, the difference between the different materials is small.

VI. BETA DOSIMETRY

There are several methods of calculation of beta dose and different applications of these methods according to the different source geometry. A point source dose rate can be calculated with the Loevinger formula [7]:

$$D'(r) = \frac{KC}{(\mu r)^2}\left\{c\left[1 - \left(\frac{\mu r}{c}\right)e^{(1-\mu r/c)}\right] + \mu r e^{(1-\mu r)}\right\} \tag{1.58}$$

where $D'(r)$ is the beta dose rate in rad per hour at distance r from the point source, r is measured in gr/cm^2, C is the source intensity in curies, c is a parameter dependent on the beta maximum energy (dimensionless), μ is the absorption coefficient in cm^2/gr, and K is a normalization constant

$$K = \frac{1.7 \times 10^5 \rho^2 \mu^3 E_{av}}{[3c^2 - e(c^2 - 1)]} \frac{\text{rad/h}}{\text{curie}} \tag{1.59}$$

where ρ is the absorber density, e is the mathematical e, and E_{av} is the beta average energy. The value of c in air is $3.11e^{-0.55E_{max}}$ and in tissue, $c = 2$ for $0.17 \le E_{max} < 0.5$ MeV, $c = 1.5$ for $0.5 \le E_{max} < 1.5$ MeV, and $c = 1$ for $1.5 \le E_{max} < 3.0$ MeV. μ in air is given by

$$\mu = \frac{16(2 - E_{av}/E^*_{av})}{(E_{max} - 0.036)^{1.4}} \text{ cm}^2/\text{gr} \tag{1.60}$$

E^*_{av} is called the hypothetical average beta energy per disintegration for a hypothetical forbidden beta disintegration having the same E_{max} as an allowed beta decay transition in the same Z element. For allowed spectra, $E_{av}/E^*_{av} = 1$.

Other simple expressions for beta dose rate calculation are available. In analogy to gamma point source dosimetry, the following equation can be used:

$$D'(r) = 2.14 \times 10^6 \rho^2 \frac{\mu}{\rho} C E_{av} \frac{e^{-\mu r/\rho}}{4\pi r^2} \frac{\text{rad}}{\text{h}} \tag{1.61}$$

for C in curies, E_{av} in MeV, ρ in gr/cm^3, $\mu/\rho = 17E_{max}^{-1.14}$ cm^2/gr, and r in gr/cm^2. Expressions for other source geometries can be found in Fitzgerald et al. [8]

VII. NEUTRON AND HEAVY PARTICLES DOSIMETRY

A. Neutron Dosimetry

Neutron dosimetry is done by transforming the number density of neutrons (or neutron flux) to dose. This is done by using the equation

$$D'(r,E) = K\phi(r,E)\left[\Sigma_s(E)\frac{AE}{(A+1)^2} + \Sigma_{n,\gamma}(E)E_\gamma B\right] \tag{1.62}$$

where $D'(r, E)$ is the dose rate in rad/h, K is a conversion factor

$$K = 5.76 \times 10^{-5} \frac{\text{rad}}{\text{h}} \Big/ \frac{\text{Mev}}{\text{cm}^3\text{s}}$$

ϕ (r, E) is the neutron flux in n/cm^2 sec, A is the atomic mass of the target nucleus, E_γ is the radioactive capture gamma-photon energy in MeV, and B is a factor representing the fraction of radioactive capture gamma-photon energy absorbed in the neighborhood of the capture. $A/(A+1)^2$ is the fraction of incident neutron energy imparted to the recoil nucleus of mass A. Σ_s is the scattering cross section and $\Sigma_{n,\gamma}$ is the (n, γ) reaction cross section. If any other reaction in addition to scattering and radiation capture takes place, the energy transferred to the substance should be included.

The energy transferred to the substance after neutron collision (first collision dose) is given by

$$D(E) = 1.602 \times 10^{-8} \sum_i \sum_j N_i \sigma_{ij}(E) \varepsilon_{ij}(E) \tag{1.63}$$

where N_i is the number of nuclei of type i per gram of substance, σ_{ij} is the cross section of the ith kind of nucleus for the reaction in which particles of type j are produced, and ε_{ij} is the average kinetic energy of the jth particle emitted by the ith nucleus.

In elastic scattering the secondary panicle is the scattered neutron, and for the isotropic case,

$$\varepsilon_{ij} = \frac{2mM_iE}{(m + M_i)^2} \tag{1.64}$$

where m is the neutron mass, M_i is the nucleus mass, and E is the neutron energy.

For an unisitropic scattering, the last expression should be multiplied by $1 - f_{li}(E)$, where $f_{li}(E)$ is given by the expansion of the elastic cross section

$$\sigma_{el}(E, \Theta) = \frac{\sigma_{el,0}(E)}{4\pi}\sum_{l=0}^{\infty}(2l+1)f_{l,i}(E)P(\cos\Theta) \tag{1.65}$$

P is the Legendre polynomial.

In the case of nuclear reaction,

$$\varepsilon_{ij}(E) = E + Q_{ij} \quad (Q_{ij} + E > 0) \tag{1.66}$$

where Q_{ij} is the reaction Q value.

B. Heavy Particles

The introduction of heavy particles (hadrons) into radiation therapy aims at improving the physical selectivity of the irradiation (e.g., proton beams) or the radiobiological differential effect (e.g., fast neutrons) or both (e.g., heavy ion beams). Each of these therapy modalities requires several types of information; absorbed dose measured in a homogeneous phantom in reference conditions; dose distribution computed at the level of the target volume(s) and the normal tissues at risk; radiation quality from which an evaluation on the RBE could be predicted; and RBE measured on biological systems or derived from clinical observation. The single beam isodoses and thus the dose distributions are similar in neutron and photon therapy. Similar algorithms can then be used for treatment planning and the same rules can be followed for dose specification for prescribing and reporting a treatment. In hadron therapy, the RBE of the different beams raises specific problems. For fast neutrons, the RBE varies within wide limits (about 2 to 5) depending on the neutron energy spectrum, dose, and biological system. For protons, the RBE values range between smaller limits (about 1.0 to 1.2). A clinical benefit is thus not expected from RBE differences. However, the proton RBE problem cannot be ignored since dose differences of about 5% can be detected clinically in some cases. The situation is most complex with heavy ions since the RBE variations, as a function of particle type and energy, dose, and biological system, are at least as large as for fast neutrons. In addition, the RBE varies with depth. Radiation quality thus has to be taken into account when prescribing and reporting a treatment. This can be done in different ways: description of the method of beam production; computed LET spectra and/or measured microdosimetric spectra at the points clinically relevant; or RBE determination. The most relevant data are those obtained for late tolerance of normal tissues at 2 Gy per fraction ('reference RBE'). Combination of microdosimetric data and experimental RBE values improves the confidence in both sets of data.

VIII. BIOLOGICAL DOSIMETRY

When dose to radiation workers or to patients is a concern, biological dosimetry is the most accurate dosimetry technique. In this way the radiation effect on the human body is measured directly without the intermediary of a technical device. No interpretation of physical or chemical phenomena taking place in the dosimeter is needed, nor is there a need for corrections.

Chromosome aberration analysis is recognized as a valuable dose-assessment method which fills a gap in dosimetric technology. Detection of chromosomal aberrations

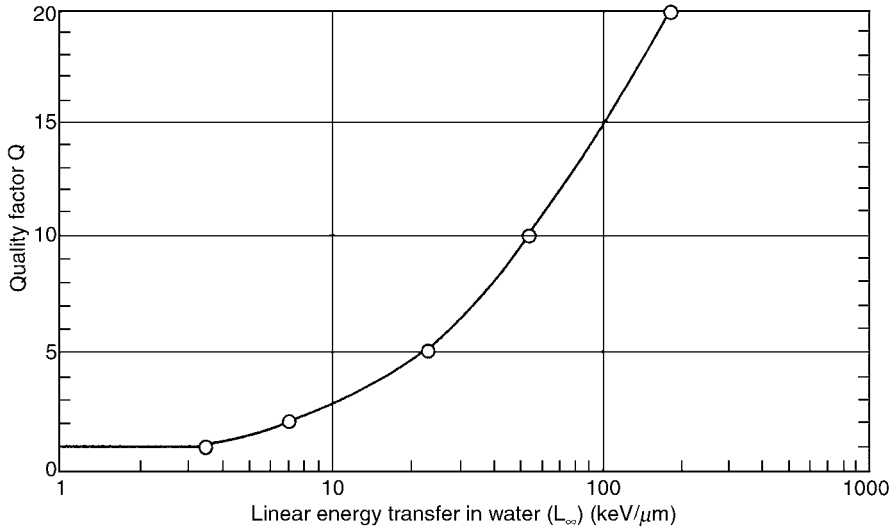

FIGURE 1.1 Quality factor as a function of linear energy transfer in water (L_∞).

in the peripheral blood lymphocytes of exposed persons is the most fully developed biological indicator of exposure to ionizing radiation. By using a distribution analysis of the aberrations, it is possible to estimate the proportion of the body exposed and the average dose absorbed by the irradiated fraction.

The influence of the microscopic distribution of the absorbed energy on the detriment is taken into account by the use of the quality factor, Q.

The ICRP [9] recommends the following approximations for the average value of Q:

- X-rays, γ-rays and electrons: 1
- Thermal neutrons: 4.6
- Other neutrons: 20
- Protons and single-charged particles of unknown energy and rest mass $\geq 1:10$
- α-particles and multiple-charged particles of unknown energy: 20

The dependence of Q on LET is shown in Figure 1.1.

The RBE varies with the LET such that a hump-shaped response curve is obtained. A generalized curve is shown in Figure 1.2.

In order to produce a dicentric aberration, DNA damage must be induced in the two unreplicated chromosomes involved such that the damaged chromosomes can undergo exchange.

As the dose increases, the contribution of two track-induced dicentrics will also increase. Thus, the dose-response curve for X-ray-induced dicentrics will be a combination of one- and two-track events, with the former being more frequent at low doses and the latter being much more frequent

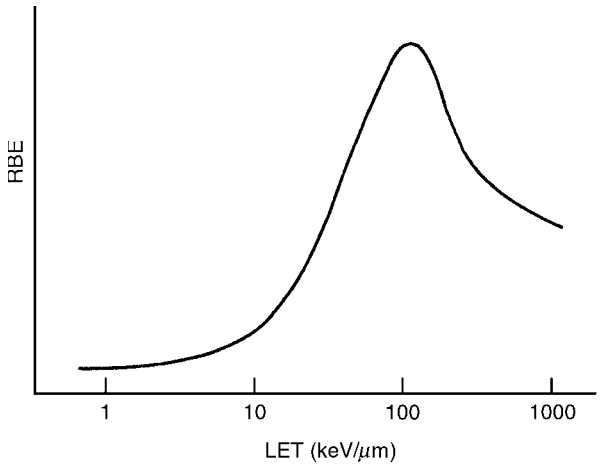

FIGURE 1.2 Generalized relationship between RBE and LET.

at high doses. The dose-response curve is generally assumed to fit the equation

$$Y = \alpha D + \beta D^2 \tag{1.67}$$

where Y is the yield of dicentrics, D is the dose, α is the linear coefficient, and β is the dose-squared coefficient.

The dose-response curve for low LET radiation (X-rays or γ-rays) will be non-linear and best fit a linear-quadratic model. The dose-response curve for high LET radiation (for example, neutrons, protons, and α-particles) will be linear, or close to linear. RBE increases with increasing LET to a maximum of 100 keV/μm and decreases at higher LET values as a result of overkill.

Figure 1.3 shows a selection of dose-response curves.

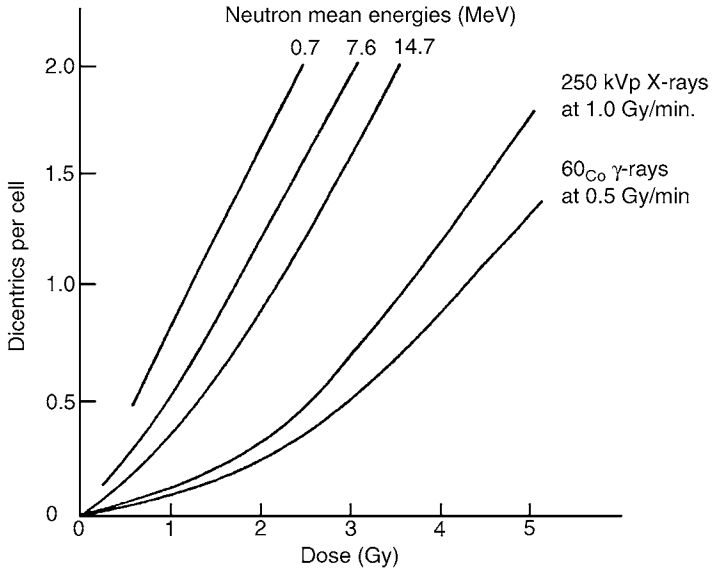

FIGURE 1.3 The relationship between dicentric yield and acute exposure to several types of radiation.