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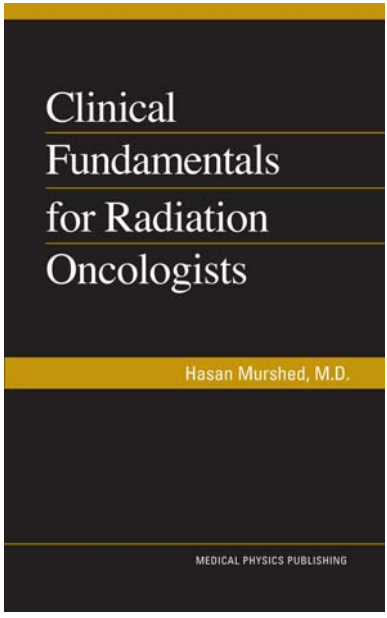

Here is a sample chapter from this book.

Clinical Fundamentals

for Radiation Oncologists

Clinical Fundamentals

for Radiation Oncologists

Hasan Murshed, M.D.

Medical Physics Publishing
Madison, Wisconsin

This volume incorporates the previous edition *Clinical Fundamentals for Radiation Oncology Residents*, © 2006 Hasan Murshed, M.D.

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DEDICATION

This book is dedicated to my children Ishraq and Ishmam.

Their faces remind me every day that “may I always act so as to preserve the finest traditions of my calling and may I long experience the joy of healing those who seek my help.”

– Hippocratic Oath

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Chapter 1

RADIATION PHYSICS, DOSIMETRY, AND TREATMENT PLANNING

Radiation has been used to treat cancer for over a century. Wilhelm Roentgen discovered x-rays on November 8, 1895, while experimenting with a gas-filled cathode tube; Henri Becquerel discovered radioactivity in 1896, while experimenting with uranium salts. It is important to know the production, physical interaction, and utilization of ionizing radiation and the basic principles of radiation physics used in the practice of radiation oncology.

THE PHYSICS OF RADIATION ONCOLOGY

Fundamental Physical Quantities

Four attributes are considered basic in the science of radiation physics: mass, energy, charge, and force.

Mass

Mass is the property of an object that measures the amount of matter within the object.

- The *rest mass* is the physical mass of an object when the object is at rest. The rest mass of an atomic particle can be converted into energy by certain nuclear processes.
- The *relativistic mass* is the mass of an object in motion relative to an observer. The relativistic mass increases as the magnitude of the velocity increases.
- For macroscopic objects, the units of mass are given in kilograms (kg), the fundamental unit of mass in the metric system.

The unit of atomic mass is the atomic mass unit (amu or u), equal to one-twelfth the mass of the most abundant isotope of carbon, ^{12}C, which is assigned a mass of exactly 12.

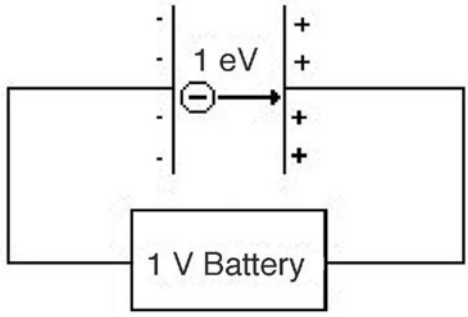


Figure 1–1. Potential energy of 1 eV in an electric field.

Energy

Energy is a measure of the ability to do mechanical work.

- *Potential energy* is the energy due to the position of the object relative to other objects and kinetic energy is energy of motion.
- The International System of Units (SI) unit of energy for macroscopic objects is the *joule (J)* and is equivalent to kg-m^2/s^2.
- The unit of energy for atomic objects is the *electron volt (eV)*. The electron volt is defined as the kinetic energy given to an electron initially at rest going through a potential difference of 1 V (see Figure 1–1).
- The following relationship exists between the atomic mass unit and the energy: 1 amu = 931 MeV.

Charge

Electric charge can be positive or negative and measures how strongly a particle is attracted to an electrical field. In classical physics, the smallest unit of negative charge is the electron (–1), while the smallest unit of positive charge is the proton (+1). The unit of electric charge is the coulomb (C).

Force

- **Coulomb Force:** The protons and electrons are held together by the coulomb force. The coulomb force predicts that if two charged particles of the same sign are brought near each other, they will cause a repulsive force between them, and if two charged particles of opposite signs are brought together, they will cause an attractive force between them, as shown in Figure 1–2. The magnitude of this force is inversely proportional to the square of the distance between the two particles. This dependence on the distance is known as the *inverse square law*.

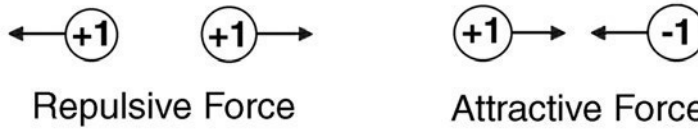


Figure 1–2. Repulsive and attractive forces between the like charged particles and the unlike charged particles, respectively.

- **Gravitational Force:** If two particles are brought near one another, they will have an attractive force between them, called the gravitational force. The gravitational force is always attractive, since the product of the masses is always positive.

Atomic Structure

The atom consists of three fundamental particles: the proton, the neutron, and the electron. The properties of these particles are listed in Table 1–1.

Atomic Models

- **Rutherford Model of the Atom:** The planetary model of the atom assumes that the protons and neutrons reside in the center of the atom (called the *nucleus*) and electrons revolve around the nucleus in circular orbits (see Figure 1–3). These electrons are said to be in shells or energy levels.
- **Bohr Model of the Atom:** The Bohr model made four assumptions to further describe the planetary model of the atom:
  (1) Electrons can only occupy certain orbits while revolving around the nucleus.
  (2) When electrons are in these stationary orbits, they do not emit radiation as predicted by classical physics.

Table 1–1. Properties of Fundamentals Particles

Particle	Charge	Mass (amu)	Rest Mass (MeV)
Proton	+1	1.007277	938
Electron	–1	0.000549	0.511
Neutron	0	1.008665	939

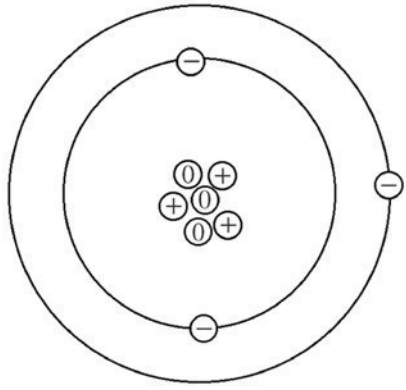


Figure 1–3. Atomic model with central nucleus consisting of protons and neutrons surrounded by orbital electrons.

(3) Each of these stationary orbits has an energy that is associated with it.
(4) Radiation is only emitted whenever an electron moves from a higher orbit to a lower orbit and radiation is absorbed whenever an electron moves from a lower orbit to a higher orbit.

- **Quantum Mechanical Forces in the Nucleus:** In addition to the electrostatic and gravitational forces discussed previously, quantum mechanics also predicts that two additional forces are present in the nucleus of the atom:
  (1) *Strong Force:* The strong force is responsible for the binding together of protons and neutrons in the nucleus and is the strongest known physical force, but it acts only over distances comparable to those between nucleons in the atomic nucleus.
  (2) *Weak Force:* The weak force is responsible for nuclear particle decay processes, such as beta decay in radioactivity.

Atomic Binding Energy

- Negative electrons are bound to the positive nucleus by the Coulomb force, an amount of energy called the *binding energy* of the atomic electron. It takes an amount of energy greater than the binding energy of that shell to remove an electron from the atom.
- Electron binding energy is a function of the radius of the electron orbit and the charge within the nucleus and will increase as the electron orbitals get closer to the nucleus and as the charge of the nucleus increases.

Table 1–2. Maximum Number of Electrons Allowed in the Atomic Shells

Shell	n	Max. e^-
K	1	2
L	2	8
M	3	18
N	4	32

Atomic Shell Filling Rules

Electron shells are labeled from the nucleus outward in two different, but related ways: (1) by letters K, L, M, N, etc., and (2) by numbers 1, 2, 3, 4, etc., (principal quantum number n) (see Table 1–2). The maximum number of electrons allowed in a given atomic shell can be determined from:

$$\text{max electrons} = 2n^2. \tag{1.1}$$

- The only exception to this shell-filling formula occurs for the outer shell of an atom. The outer shell of an atom can contain no more than eight electrons.
- When an electron acquires enough energy from an incident photon to leave the inner orbit of the atom, a vacancy will be created in that shell. An electron from one of the outer energy shells then promptly fill this vacancy, emitting the excess energy as a photon in the process. Photon thus emitted are known as *characteristic x-rays.*
- However, there is a competing process to characteristic radiation. After an electron transitions from an outer shell to an inner shell, instead of the atom emitting a characteristic photon, the atom can then absorb this energy and enter into a higher, excited state. This excess energy is subsequently emitted from the atom by the ejection of one or more electrons. This process of ejecting an electron from the atom's shells is called the *Auger effect* and the electron that is released is called an *Auger electron.*

Nuclear Structure

Atoms are identified by their atomic symbols ${}^{A}_{Z}\mathrm{X}$, where X is the atomic symbol, A is the mass number (number of protons and neutrons), and Z is the atomic number (number of protons or number of electrons). Note that the number of neutrons N in an atom can be determined by:

$$N = A - Z. \tag{1.2}$$

Nuclear Binding Energy

The particles contained in the nucleus are bound together by the strong and the weak nuclear forces, as discussed previously. If one compares the mass of a nuclide with the mass of its constituent components, it is found that the mass of the nucleus is always less than the mass of the constituent components. This deficiency of mass is called the *mass defect,* and the energy required to separate the nucleus into its constituent particles is called the *binding energy,* E_b, of the nucleus, and is given by Einstein's famous mass-energy equivalence relationship:

$$E_b = \Delta mc^2, \tag{1.3}$$

where Δm is the mass difference of the nucleus and its constituent particles and c is the speed of light (=1 when dealing in nuclear units). **Note:** Nuclear processes that cause energy to be released will always increase the binding energy of the resultant nucleus due to the conversion of mass into energy.

RADIOACTIVE DECAY

Radioactivity is the process by which an unstable nucleus transforms by giving off the excess energy and forming a new stable element. The transformation may involve the emission of electromagnetic radiation or emission of particles, involving mechanisms such as beta decay, alpha decay, or isomeric transitions. Examining the ratio of neutrons to protons in all stable nuclei, the following conclusions can be made (see Figure 1–4):

- If *Z* is less than or equal to 20, the ratio of neutrons to protons is 1.
- If is *Z* greater than 20, the ratio becomes greater than 1 and increases with *Z*.

As more protons are added to the nucleus, the effects of the coulomb force begin to overwhelm the strong nuclear forces, which can make an atom unstable. This unstable nucleus will tend to lose energy by different decay mechanisms, described in the next section, to reach a more stable state.

Special types of nuclei are defined as follows:

- **Isotopes:** Isotopes are nuclides that have the same number of protons (*Z*) and a different number of neutrons (*A* – *Z*). Examples of two isotopes are ${}^{5}_{3}\mathrm{Li}$ and ${}^{6}_{3}\mathrm{Li}$.
- **Isobars:** Isobars are nuclides that have the same number of total particles in the nucleus (*A*) and a different number of protons (*Z*) and neutrons (*A* – *Z*). Example of isobars are ${}^{40}_{19}\mathrm{K}$ and ${}^{40}_{20}\mathrm{Ca}$.
- **Isotones:** Isotones are nuclides that have the same number of neutrons (*A* – *Z*) and a different number of protons (*Z*). Examples of isotones are ${}^{14}_{6}\mathrm{C}$ and ${}^{15}_{7}\mathrm{N}$.

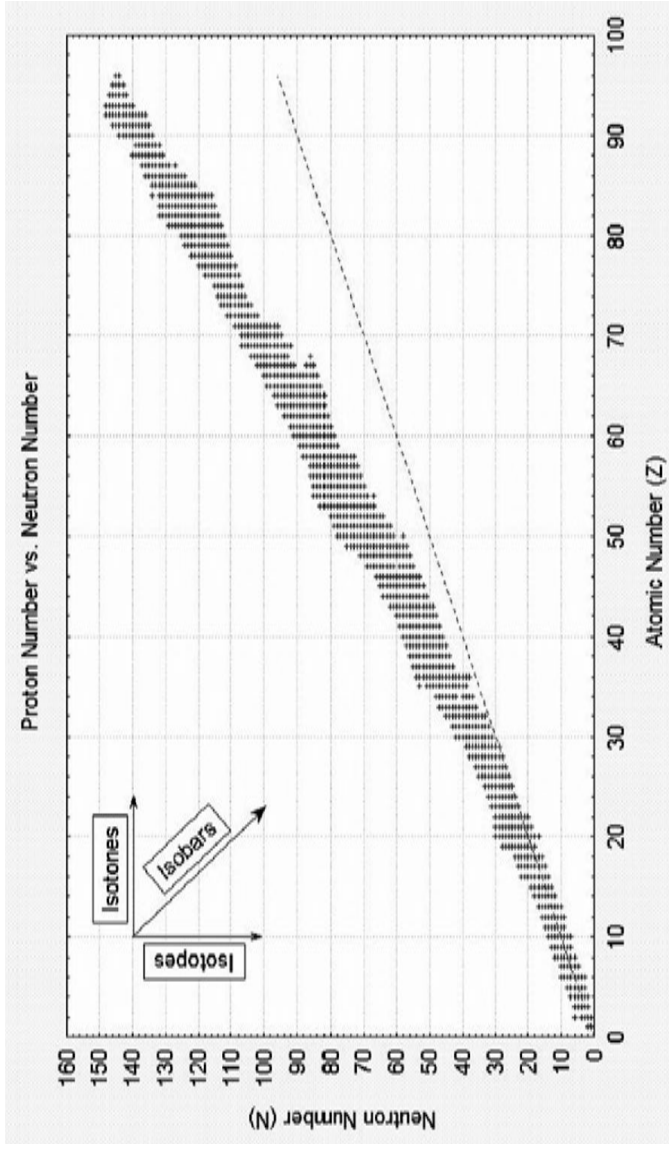


Figure I–4. Neutron vs. proton in stable nuclei.

- **Isomers:** Isomers are nuclides that have the same number of total particles in the nucleus (*A*) and the same number of protons (*Z*) but different levels of energy in the nucleus. Examples of isomers are $^{99}_{43}\mathrm{Tc}$ and $^{99m}_{43}\mathrm{Tc}$.

Modes of Radioactive Decay

Beta Decay

Beta decay is the process by which a radioactive nucleus ejects either a negatively charged electron or a positively charged positron. There are two types of beta decay: β^- (beta minus or negatron) and β^+ (beta plus or positron).

- If a radionuclide has a high number of neutrons (high n/p ratio), it tends to reduce the n/p ratio by converting a neutron into a proton, negatron, and antineutrino.
- If a radionuclide has a deficit of neutrons (low n/p ratio), it tends to increase the n/p ratio, either converting a proton into a neutron and a positron (positron emission) or by capturing an orbital electron (electron capture).
- In positron emission, a proton is converted into a neutron, positron, and a neutrino. The creation of a positron requires 1.02 MeV of energy to be available from the nuclear decay. The positron is an unstable antiparticle of the electron, possessing the same mass but opposite charge. Once the traveling positron has slowed down enough, the positron and another electron will annihilate each other, and their rest masses are converted into energy (1.02 MeV). This energy appears as two 0.511 MeV annihilation photons traveling in opposite directions (see Figure 1–5).
- However, there is a competing process to positron emission. An orbital electron can be captured by the nucleus. This process is called the *electron capture.* The nucleus then rearranges its nuclear structure and transforms a proton into a neutron in order to reach a stable state. Since the orbital electron is captured by the nucleus, a hole will exist in that orbital electron and a higher orbital electron will fill this hole, and characteristic x-rays and Auger electrons will be emitted from the atom due to this shell filling.

Alpha Decay: Radionuclide that has a high *Z* (defined to be greater than 82) will decay most frequently by the emission of a helium nucleus, or alpha particle (α). The alpha particle has the same nuclear structure as the nucleus of a helium atom, and is sometimes written as $^4_2\mathrm{He}$.

Isomeric Transitions: The most stable arrangement of the nucleus in an atom is called the *ground state*. In some nuclear decays, the daughter nucleus stays in a higher excited state for some period of time; this is said to be a *metastable state.* The only difference between the metastable state and the final ground state is a difference in energy, hence the two states

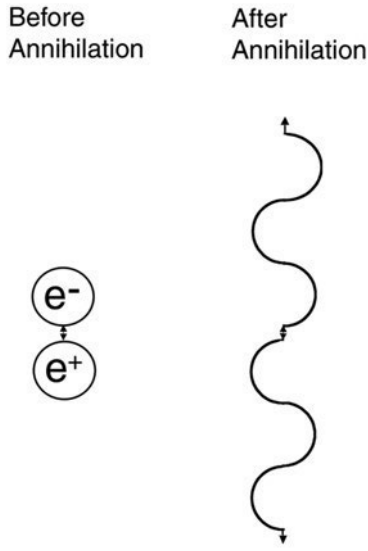


Figure 1–5. The conversion of mass into energy; the annihilation radiation is traveling in the opposite direction.

are called *isomers* and the transition from the metastable state to the ground state is called an *isomeric transition.* There are two competing methods by which a nucleus can lose excess energy during an isomeric transition: gamma emission and internal conversion.

- In *gamma emission* the nucleus can release the excess energy by the direct emission of one or more gamma rays from the nucleus.
- In *internal conversion* the nucleus can release its excess energy by the emission of one or more of the orbital electrons from the atom. If an inner shell electron is removed from the atom, shell filling will occur and will result in characteristic x-rays and/or Auger electrons being released, as discussed previously.

Mathematics of Radioactive Decay

The mathematics of radioactive decay depends on the observation that in a large collection of N radioactive atoms, the number of decays ΔN that occur in a time interval Δt is found to be proportional to Δt and to the total number N of radioactive nuclei. The proportionality constant λ is called the *decay* (or disintegration) *constant* and gives the rate at which a radionuclide is disintegrating per unit time. The units of the decay constant are s^{-1} (or disintegrations per second).

If we solve the differential equation:

$$\Delta N = -\lambda \times N \times \Delta t \tag{1.4}$$

The final result is an equation that allows us to calculate the number of radioactive atoms N(t) at any time t:

$$N(t) = N_0 e^{-\lambda t}, \tag{1.5}$$

where N(t) is the number of atoms remaining at time t, N_0 is the number atoms at time t=0, e is the mathematical constant 2.718, λ is the disintegration constant, and t is the elapsed time. (See Figure 1–6.)

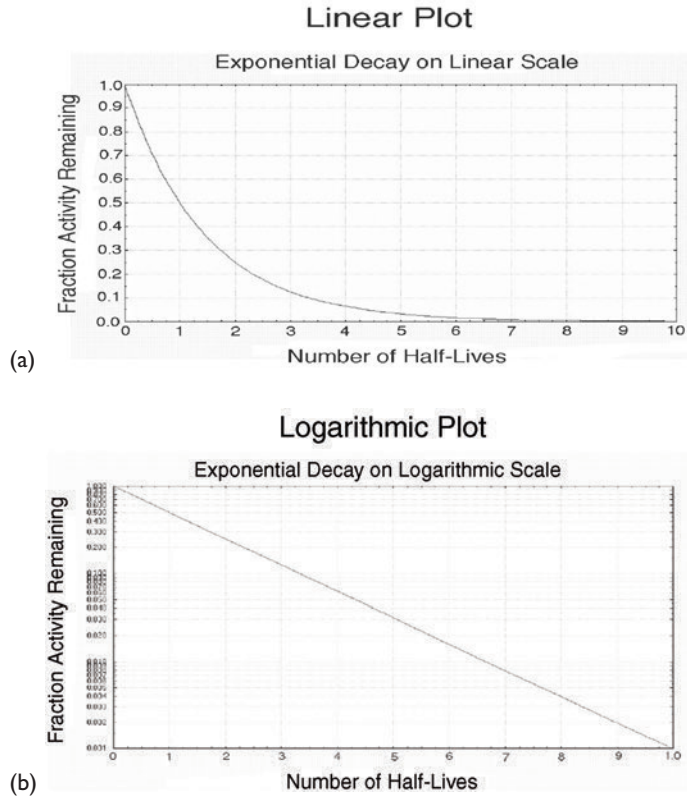


Figure 1–6. Graphs of the equation for exponential decay on (a) linear and (b) semilogarithmic paper.

- **Activity:** Activity is the rate of decay of a radioactive material at any given point in time. The activity (A) is related to the number of disintegrations by:

$$A(t) = -\lambda \cdot N(t). \tag{1.6}$$

The equation for N(t) can be rewritten in terms of activity A(t):

$$A(t) = A_0 e^{-\lambda t}, \tag{1.7}$$

where A(t) is the activity remaining at time t, A_0 is the activity at time t=0, e is the mathematical constant 2.718, λ is the disintegration constant, and t is the elapsed time.

- **Units:** The original unit of activity was the curie (Ci). The curie is defined as the number of disintegrations given off by 1 gram of radium (^{226}Ra) and is equal to 3.7×10^{10} disintegrations per second (dps). The newer SI unit of activity is the becquerel (Bq), defined by:

$$1 \text{ Bq} = 1 \text{ dps}. \tag{1.8}$$

The relationship between the curie and the becquerel is then given by:

$$1 \text{ Ci} = 3.7 \times 10^{10} \text{ Bq}. \tag{1.9}$$

- **Half-Life:** The half-life of a radionuclide is the time that it takes for the radionuclide to decay to half of its original value. It is denoted by $t_{1/2}$, and can be calculated from the disintegration constant by:

$$t_{1/2} = \frac{0.693}{\lambda}. \tag{1.10}$$

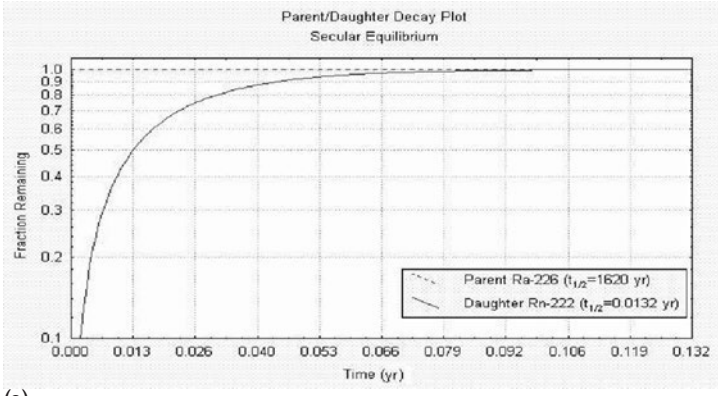
Note that the following relationship exists between the average or mean life of a radionuclide and the half life of a radionuclide:

$$t_{avg} = 1.44 \times t_{1/2}. \tag{1.11}$$

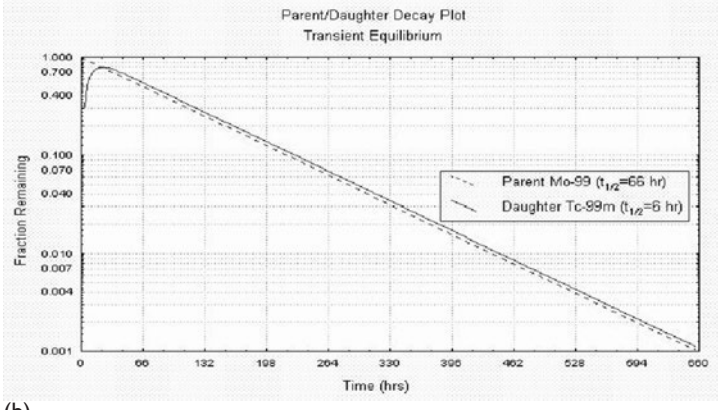
Equilibrium

As radioactive material decays, there can be situations that arise where the radioactive parent nucleus, with a decay constant λ_p, decays to an intermediate daughter state that is also radioactive. This daughter state will also decay with some decay constant, λ_d. Depending on how long

the half-lives of the parent and the daughter are in relation to each other, two different equilibrium situations can occur. This correlation of activity of the parent and the daughter after some period of time is called *equilibrium* (see Figure 1–7).



(a)



(b)

Figure 1–7. (a) A secular equilibrium plot showing the activity of the parent and daughter for different times t. (b) A transient equilibrium plot showing the activity of the parent and daughter for different times t.

- **Secular Equilibrium:** If the daughter half-life is much shorter than the parent half-life, then after some period of time, a state of secular equilibrium is reached. An example of secular equilibrium is the decay of ^{226}Ra:

$$^{226}\text{Ra}(t_{1/2} = 1620\ \text{y}) \rightarrow {}^{222}\text{Rn}(t_{1/2} = 4.8\ \text{d}). \qquad (1.12)$$

  Secular equilibrium is characterized by a gradual buildup of activity of the daughter until it reaches and is equal to the level of the parent.

- **Transient Equilibrium:** If the daughter half-life is not much shorter than the parent half-life, then after some period of time a state of transient equilibrium is reached. An example of transient equilibrium is given by the decay of ^{99}Mo to ^{99m}Tc:

$$^{99}\text{Mo}(t_{1/2} = 66\ \text{h}) \rightarrow {}^{99m}\text{Tc}(t_{1/2} = 6\ \text{h}). \qquad (1.13)$$

  This type of equilibrium is characterized by a gradual buildup of the activity of the daughter until it eventually equals and then slightly exceeds the activity of the parent. After that point the activity of the daughter follows the activity of the parent, always exceeding the parent by a small amount.

ELECTROMAGNETIC RADIATION AND PROPERTIES OF INTERACTION

X- and gamma radiation are a part of a larger set of photon radiation called the *electromagnetic spectrum*. Figure 1–8 shows the types of radiation in the this spectrum.

- All types of photons have two fundamental parameters:
  - **Wavelength λ:** The wavelength of a photon is the distance from one point on the photon wave to the same exact point on the next repetition of the wave, such as the distance from the crest of the wave to the crest of the next repeat of the wave (see Figure 1–9).
  - **Frequency υ:** The frequency of a photon is the number of times the wave oscillates per second.
- Photons are known to travel at a constant velocity in space. This velocity, denoted by c, is called the *speed of light*. It has a value of 3×10^8 meters per second (m/s) in a vacuum The relationship between the frequency υ of a photon and its wavelength λ, is given by:

$$\nu = \frac{c}{\lambda}, \qquad (1.14)$$

  where υ is the frequency of the photon, λ is the wavelength of the photon, and c is the speed of light.

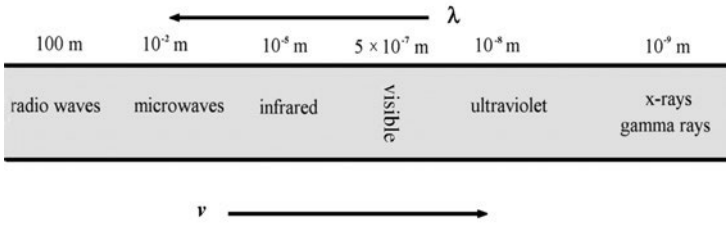


Figure 1–8. The electromagnetic spectrum. (Reprinted from ref 1, McDermott PN, Orton CG, *The Physics & Technology of Radiation Therapy,* Fig. 2.13, p. 2-18, © 2010, with permission from the author.)

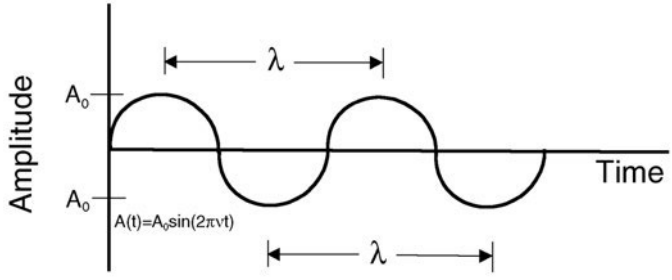


Figure 1–9. Illustration of an electromagnetic wave showing the wavelength and frequency.

- Photons are known to have zero mass and zero charge, but do have energy given by:

$$E = h\upsilon, \tag{1.15}$$

where υ is the frequency of the photon (in units of s^{-1} or Hz [hertz]), and h is the physical constant called Planck's constant (6.62×10^{-34} J-s).

- The energy of a photon can be rewritten in terms of wavelength by:

$$E = \frac{hc}{\lambda}. \tag{1.16}$$

Photon Interaction Process, Kerma, Absorbed Dose

As photons travel through a material, they release atomic orbital electrons in their path. These freed electrons, called the *primary electrons*, cause most of the energy deposition that occurs in therapeutic photon beams. Photon interactions with matter are a two-step process:

- **Energy Transfer and Kerma:** An incident photon will interact with an atomic orbital electron in the atom of the material through some photon interaction mechanism (such as photoelectric, Compton, or pair production, which will be described later). During these interactions, photons will transfer some or all of their kinetic energy to an atomic orbital electron. This orbital electron can gain enough kinetic energy to leave the atomic orbit and become a primary electron. This process of energy transfer from uncharged particles (photons) to charged particles (electrons) is called the *kerma* (*k*inetic *e*nergy *r*eleased in *ma*tter). The SI unit for kerma is the joule per kilogram (J/kg), renamed the gray (Gy).
- **Energy Absorption and Absorbed Dose:** The primary electrons that were released in the energy transfer process will then interact with other atomic orbital electrons and release ions and excited atoms along their irregular tracks. In addition, primary electrons with enough energy may cause secondary tracks of their own; these electrons are called *delta rays*. The electrons that deposit their energy at the interaction site give rise to an important quantity called the *absorbed dose*. Absorbed dose is the energy absorbed in a material per unit mass.
- The old unit of absorbed dose is the rad (radiation absorbed dose). Rads have been replaced in the SI system with the gray (Gy). The following relationship exists between the rad and the Gy:

$$100 \text{ rad} = 1 \text{ J/kg} = 1 \text{ Gy}. \tag{1.17}$$

- Because the gray is so large compared to the rad, a subunit is defined called the centigray (cGy). The equivalence of the rad and cGy is:

$$1 \text{ rad} = 1 \text{ cGy}. \tag{1.18}$$

Modes of Photon Interaction

The interactions mechanisms for photons are described below. These interactions transfer energy from the radiation to the irradiated materials, where it is ultimately dissipated as heat.

- **Coherent Scattering:** Coherent scattering is dominant in the low-energy region (up to 50 keV) and high Z materials. A photon interacting with the atom may be absorbed by the electrons in the atom causing them to oscillate. These electrons will then reradiate this excess energy out of the atom by emitting a photon with the same energy as the incident photon, but at a different angle. Note that in this type of interaction, no energy is transferred to the atom.
- **Photoelectric Absorption:** Photoelectric absorption is a photon interaction in which an incident photon is totally absorbed by an inner shell electron (see Figure 1–10a). If this electron, called a *photoelectron,* gains enough energy to overcome its binding energy, it will then eject from the atom's shell. This is the main interaction responsible for diagnostic imaging.
- **Compton Scattering:** In the Compton interaction the incident photon interacts with an outer shell electron (see Figure 1–10b). The incident photon is reduced in energy and scattered at some angle θ. The outer shell electron absorbs a fraction of the incident photon's energy and is emitted at an angle ϕ with energy equal to the difference between the incident photon and the scattered photon.
- There are some special cases of Compton interactions, as described below:
  - *0° Photon Scatter (Grazing Hit):* If the incoming photon makes a grazing hit with an atomic electron, then the scattered photon will receive maximum energy and the scattered electron will receive minimum energy.
  - *90° Photon Scatter:* If the incoming photon scatters 90° from the incident photon's direction, then the scattered photon will always have an energy of 0.511 MeV. These photons are important in shielding calculations.
  - *180° Photon Scatter (Direct Hit):* If the incoming photon scatters 180° from the incident photon's direction, with respect to the beam direction, then the scattered photon will always have an energy of 0.255 MeV.
- **Pair Production:** In a pair production interaction the incident photon may interact with the atomic field around the nucleus, causing the photon to lose all of its energy (see Figure 1–10c). This energy is then converted into matter in the form of an electron-positron pair. Positron is antimatter and as it slows down while traveling in matter, it combines with an electron to give rise to two annihilation photons, each having the energy of 0.51 MeV. The pair production interaction cannot occur unless the incident photon has more than 1.02 MeV energy.
- **Photodisintegration:** The photonuclear interaction occurs when a high-energy photon is absorbed by the nucleus. If an interaction occurs, the nucleus may then eject a neutron, proton, or alpha particle and cause a breakup of the nucleus, called *photodisintegration.* This interaction begins to occur at 10 MV, but becomes increasingly important when linear accelerators operate at energies above 15 MV.

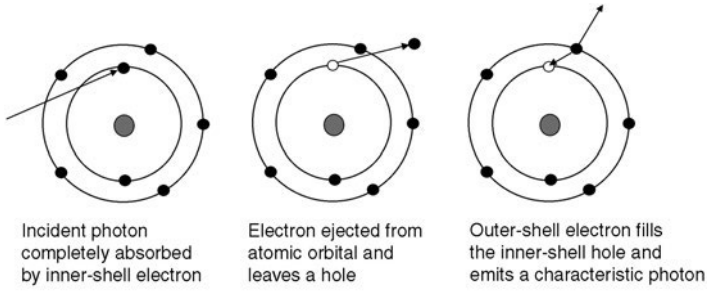


Figure 1–10a. Photon interaction of photoelectric absorption.

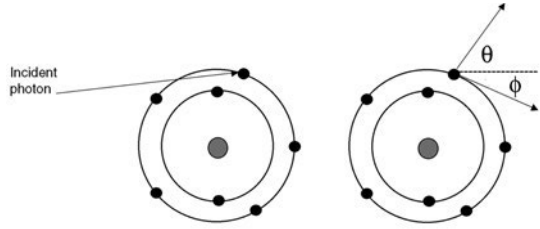


Figure 1–10b. Photon interaction of Compton scattering.

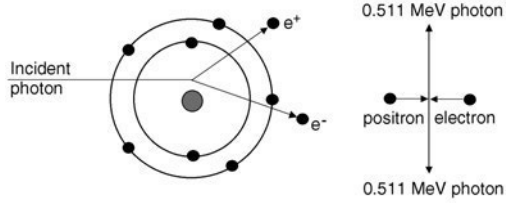


Figure 1–10c. Photon interaction of pair production.

At this energy, neutrons begin to be produced as a result of photodisintegration of atoms in the metals of the collimator, flattening filter, target, and with the oxygen and other gaseous components in the air.

- **Atomic Number Dependence of Interaction Probabilities:**
  - Coherent scattering: Probability of interaction is a function of Z
  - Photoelectric absorption: Probability of interaction is a function of Z^3
  - Compton scattering: Probability of interaction is independent of Z
  - Pair production: Probability of interaction is a function of Z^2

Table 1–3. Probability of a Given Interaction as a Function of Photon Energy*

Photon Energy (MeV)	Interactions (%)		
	P_{photo}	P_{Comp}	P_{pair}
0.010	92	3	0
0.030	36	50	0
0.150	0	100	0
4.0	0	94	6
10.0	0	77	23
30.0	0	43	57
100.0	0	16	84

*Data from ref 2, Johns HE, Cunningham JR, *The Physics of Radiology,* 4th edition, Table 5-5, p. 163, 1983.

- **Energy Dependence of Interaction Probabilities:** Table 1–3 shows the probability of a given interaction occurring as a function of energy in water.

Mathematics of Photon Attenuation

The process of photon attenuation is a random event and it must be treated statistically. The mathematics of photon attenuation depends on the assumption that as if N photons pass through a material of thickness x, a given fraction μ of these photons will be removed from the beam. Stated in mathematical terms:

$$\frac{\Delta N}{\Delta x} = -\mu N, \tag{1.19}$$

where μ is a constant called the *linear attenuation coefficient.*

The linear attenuation coefficient μ gives the rate at which photons are removed from a material per unit thickness. The units of the linear attenuation coefficient are cm^{-1}. If we integrate this differential equation, we arrive at a formula that can be used to calculate the number of photons that will remain after a photon beam has passed through a thickness x of material, given by:

$$N(x) = N_0 e^{-\mu x}, \tag{1.20}$$

where N(x) is the number of photons that are left after going through material of thickness x of material, N_0 is the number of photons that would be left if no material were present, x is the thickness of absorber material (units of cm), and μ is the linear attenuation coefficient.

This type of attenuation is known as *exponential attenuation.* It is known that the amount of attenuation for the same thickness x of material is different for different density materials; it follows that μ depends on the density of the material. If we divide μ by ρ, we obtain the *mass attenuation coefficient,* (μ/ρ). The mass attenuation coefficient has units of cm^2/g and is independent of density.

PARTICULATE RADIATION PROPERTIES AND INTERACTIONS

Particulate radiation is generally classified as particles that possess mass. The interaction of the particulate radiation with matter is different from the interaction of photon with the matter. This is mainly because the particulate radiation has mass and most of the time also possesses electrical charge. The interactions of the electron, neutron, and the heavy particles are described below.

Interactions of Electrons

Electrons differ from heavy charged particles by the fact that an electron can give up a large portion of its kinetic energy in a single interaction. Electrons do not possess the usual Bragg peak that is associated with heavier charged particles that have mass and charge because electrons readily change direction after each interaction.

- **Interactions with Atomic Electrons:** Free electrons interact with other atomic electrons predominantly by inelastic collisions, causing excitation and ionization of the atom. *Excitation* is the promotion of an orbital electron to a higher energy level in the atom, without the ejection of the orbital electron. On the other hand, if the incident electron had enough energy to eject the orbital electron, then the interaction is called an *ionization* (see Figure 1–11).
- **Interactions with Nuclei (Bremsstrahlung):** Electrons with kinetic energy will interact with atomic nuclei, mostly by inelastic collisions. As the electron is slightly bent from its original path by the Coulomb force of the atomic nucleus, it decelerates and loses energy in the form of a photon. This photon is called a *bremsstrahlung photon* (see Figure 1–12).

Interactions of Neutrons

Neutron interactions can produce a wide range of subatomic particles, recoil nuclei, and photons. These wide arrays of particles that are produced can deposit their energy in very different ways, which will produce

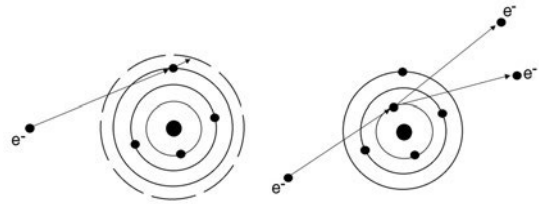


Figure 1–11. Excitation and ionization of an atom.

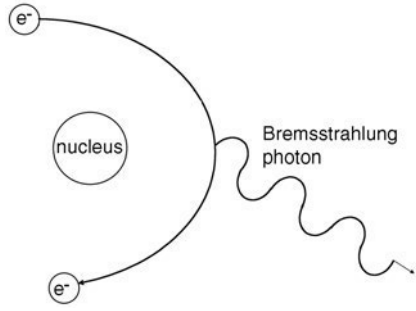


Figure 1–12. Creation of bremsstrahlung radiation.

differing biological effects. There are five types of neutron interactions that can occur when a neutron encounters a nucleus.

- **Elastic Scattering (n, n):** The simplest neutron interaction is for a neutron to elastically collide with the nucleus of an atom, which is denoted as (n, n). After the collision, the neutron has lost some energy and is deflected at some angle. The lost energy is transferred to the recoiling nucleus.
- **Inelastic Scattering (n, nγ):** Another neutron interaction is for a neutron to be momentarily absorbed by a nucleus, leaving the nucleus in an excited state. This unstable situation may result in the ejection of another nuclear particle and a γ-ray.
- **Transmutation (n, p), (n, α):** In a transmutation reaction, the nucleus may completely absorb a neutron and form an unstable compound nucleus. This excited nucleus may then re-emit a charged particle, such as (n, p), (n, d), or (n, α), and transmute to a different element.
- **Radiative Capture (n, γ):** The process of radiative capture is one of the most common nuclear reactions. In this process, the nucleus completely absorbs the neutron and emits a γ-ray photon.

- **Fission (n, f):** In the fission process, a neutron is absorbed by a heavy nucleus like $^{235}_{92}U$, which subsequently breaks apart into two lighter nuclei.

Interactions of Heavy Charged Particles

Heavy charged particles generally include any charged particle with rest mass above a proton. These include protons, heavy ions, and negative pions. A heavy charged particle can transfer only a small amount of its energy in a single collision, because of its large mass, and does not change direction appreciably while traveling through matter. Heavy charged particles with large amounts of kinetic energy generally interact with matter by undergoing inelastic collisions with atomic electrons. In this process they give up a portion of their energy at each interaction (without changing direction appreciably), finally giving up a large portion of energy at the end of their range, causing the *Bragg peak.*

THE PHYSICS OF DOSIMETRY

The radiation dose prescribed by a radiation oncologist for a patient needs to be quantified by a dosimetrist or a medical physicist to determine treatment machine settings, known as a monitor unit (MU). It is important to understand the basic parameters used for dosimetry calculation and treatment planning. The following paragraphs provide a brief review of the concepts in dose calculation leading to MU settings.

Inverse Square Law

The inverse square law can be derived by considering a point source that is emitting radiation equally in all directions; two spheres are centered on this source with radii r_1 and r_2 ($r_2 > r_1$) (see Figure 1–13).

The intensity I of the photons emitted from the source is defined as the number N of photons going through a spherical surface divided by the area A of the sphere:

$$I_1 = \frac{N_1}{A_1} \tag{1.21a}$$

and

$$I_2 = \frac{N_2}{A_2}. \tag{1.21b}$$

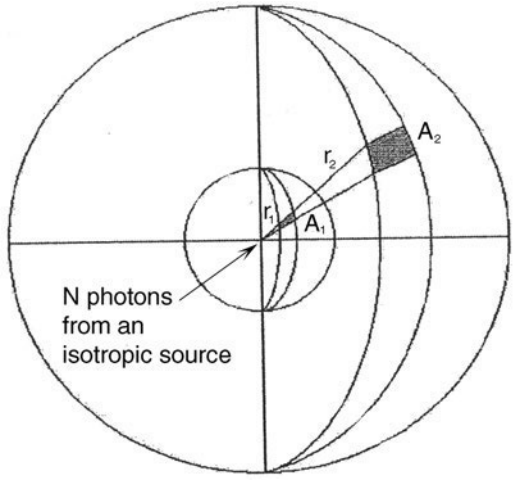


Figure 1–13. The inverse square law: r_1 is the distance from the point source to a sphere of area A_1, and r_2 is the distance from the point source to a sphere of A_2.

It is assumed that no photons are lost due to absorption as they travel away from the source, so:

$$N_1 = N_2. \tag{1.22}$$

Noting that the area of a sphere is $A = \pi r^2$; solving these equations yields:

$$I_1\left(\pi r_1^2\right) = I_2\left(\pi r_2^2\right) \tag{1.23a}$$

or

$$\frac{I_2}{I_1} = \left(\frac{r_1}{r_2}\right)^2. \tag{1.23b}$$

As can be seen from equations (1.23), the ratio of intensities of a photon beam is inversely proportional to the square of the ratio of the distances from the source. This decrease in the number of photons at a function of distance is known as the *inverse square law.*

Backscatter Factor (BSF)

Consider the following two experimental arrangements. In Figure 1–14, situation (a) illustrates an exposure calibration of a photon beam in air. Situation (b) illustrates the irradiation conditions of the phantom. The

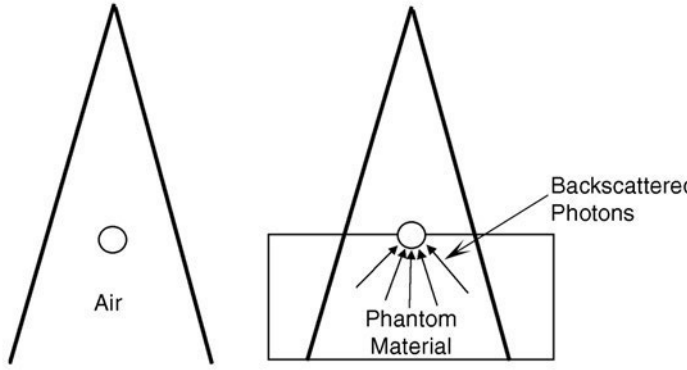


Figure 1–14. Backscatter factor measurement: (a) dose measurement in air; (b) dose measurement in phantom surface at the same point.

reading on the electrometer in (b) is increased considerably compared to (a). This increase in dose, caused by radiation that was scattered back toward the probe from the phantom/patient is called *backscattered radiation*. The factor used to correct for this effect is called the *backscatter factor (BSF)*. The backscatter factor is defined as:

$$\text{BSF} = \frac{\text{Exposure at phantom surface}}{\text{Exposure at same point with no phantom present}}. \quad (1.24)$$

The BSF increases as the energy and the field size increases; however it is independent of SSD (source-to-surface distance).

Peak Scatter Factor (PSF)

The backscatter factor only applies to low-energy photon radiation. With the use of megavoltage radiation, a build-up effect occurred that shifts the depth maximum dose, d_{max}, from the surface to a new depth. A quantity similar to the BSF was defined and called the peak scatter factor (PSF) (see Figure 1–15):

$$\text{PSF} = \frac{\text{Dose in phantom at } d_{max}}{\text{Dose in free space at same point}}, \quad (1.25)$$

where d_{max} is the maximum dose and dose in free space is the dose to an ionization chamber with a small amount of build-up material wrapped around it to provide electronic equilibrium.

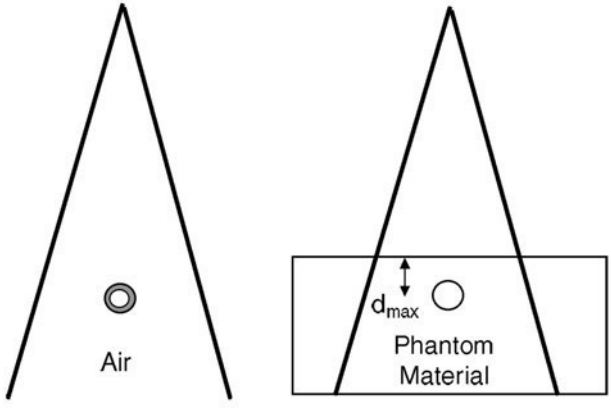


Figure 1–15. Peak scatter measurement. The build-up cap placed around the chamber in air measurement is to give dose in free space.

Depth of Maximum Dose (d_{max})

The depth of maximum dose, d_{max}, is the depth at which buildup ends. It is at this point that electronic equilibrium is most closely established.

- This buildup depth, d_{max}, is a function of energy as seen in Table 1–4, as the nominal energy increases the d_{max} depth increases.

Table 1–4. Depth of d_{max} Varies with Nominal Beam Energy (10×10 cm^2)*

Nominal Energy	d_{max} (cm)
Co-60	0.5
4 MV	1.0
6 MV	1.5
10 MV	2.5
15 MV	3.0
20 MV	3.5
25 MV	4.0
34 MV	5.0

*Data from ref 3, Jani SK, *Handbook of Dosimetry Data for Radiotherapy,* Table 1.I1, p. 63, 1993.

Table 1–5. Depth of d_{max} (cm) Varies with Field Sizes*

X-Ray Beam Energies; SSD = 100 cm					
Nominal Energy	Field Size (cm)				
	5×5	10×10	15×15	20×20	30×30
Co-60	0.5	0.5	0.5	0.5	0.5
4 MV	1.0	1.0	1.0	0.90	0.90
6 MV	1.6	1.5	1.5	1.4	1.4
10 MV	2.4	2.3	2.1	1.9	1.8
14 MV	2.9	2.7	2.5	2.4	2.3

*Data from ref 3, Jani SK, *Handbook of Dosimetry Data for Radiotherapy,* Table 1.I1, p. 63, 1993.

- Note that d_{max} also varies with field size and shifts toward the surface with increasing field size, which is more pronounced with higher-energy photons, as shown in Table 1–5.

Percentage Depth Dose (PDD)

One of the most used quantities in dosimetry is the dose at any depth along the central axis of the radiation beam. This quantity is called *percentage depth dose (PDD)* (see Figures 1–16 and 1–17). The PDD is defined by:

$$\text{PDD} = \frac{\text{Dose at depth along central axis}}{\text{Dose at depth of maximum dose along central axis}}. \quad (1.26)$$

The PDD is a function of the energy of the beam, the depth of measurement, the field size of the beam, and the SSD.

- The PDD is dependent on the following parameters:
  - **Energy:** As the beam energy increases, the PDD (for a fixed depth) increases. This is due to the greater penetrating power of the photon beam.
  - **Depth:** As the depth of measurement increases, the PDD decreases (with the exception of the build-up region). This is due to the exponential attenuation of the photon beam as it passes through the patient.

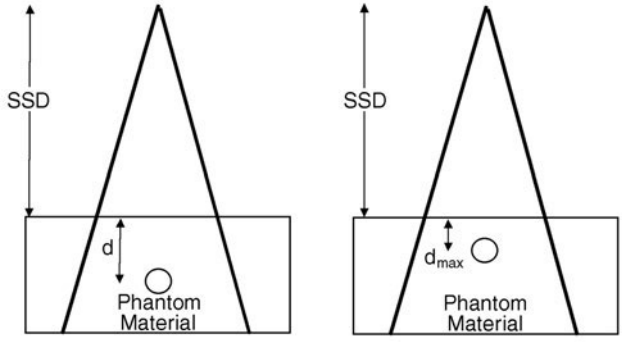


Figure 1–16. Percent depth dose (PDD) measurement: (a) dose measured at depth of central axis; (b) dose measured at d_{max} along central axis.

- **Field Size:** As the field size increases, the PDD increases. This is due to the increased scatter due to the larger collimator and patient area irradiated.
- **SSD:** As the SSD increases, the PDD increases. This is due to the definition of the PDD and the inverse-square law.

• **Mayneord F-factor:** PDD tables are usually measured at the calibration SSD (usually 80 or 100 cm). It is sometimes necessary to increase the SSD to treat a field larger than the machine is capable of treating. Since PDD is dependent on SSD, the PDD table will not be correct for the new SSD. The PDD at the new SSD can be calculated by a correction factor called the Mayneord F-factor (not to be confused with the roentgen-to-rad conversion f-factor) (see Figure 1–18). If SSD_1 denotes the source-to-surface distance of the PDD table that you have measured, d is the depth, d_{max} is the depth of maximum dose, and SSD_2 denotes the SSD to which you want to correct the PDD, the F-factor can be calculated by:

$$F = \left[\left(\frac{SSD_2 + d_{max}}{SSD_1 + d_{max}} \right) \left(\frac{SSD_1 + d}{SSD_2 + d} \right) \right]^2 . \quad (1.27)$$

The corrected PDD at the new SSD is then given by:

$$PDD_{NewSSD} = PDD_{OldSSD} \cdot F, \quad (1.28)$$

where PDD_{oldSSD} is the PDD at the original SSD.

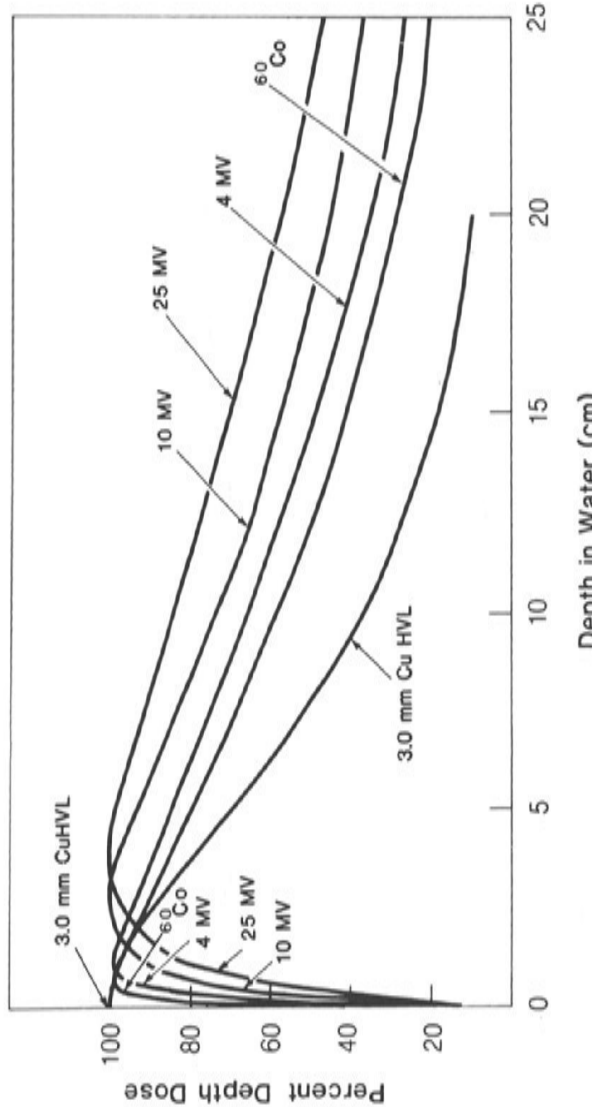


Figure 1–17. Central axis percent depth dose for different energies. (Reprinted from ref 4, Khan F, *The Physics of Radiation Therapy*, 3rd edition, Fig. 9.3, p. 163, © 2003 with permission from Lippincott Williams and Wilkins.)

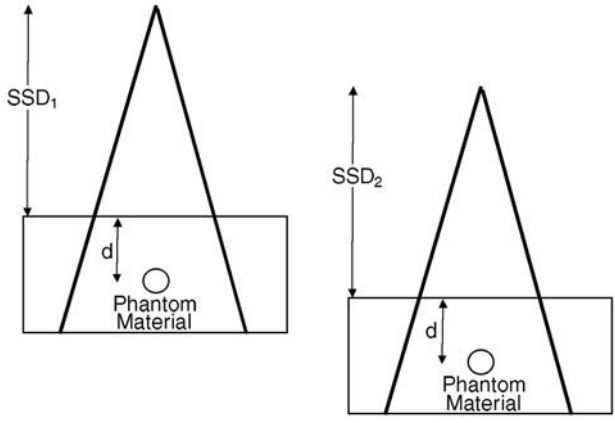


Figure 1–18. Diagram showing Mayneord F-factor calculation.

- **Equivalent Squares:** For a given rectangular field, a square field can be found that is equivalent with respect to some dosimetric parameter; i.e., it had equal PDD, BSF, or output factor. This square is called an *equivalent square field*. The equivalent square field calculation is useful because the patient's radiation treatment fields are often irregular in shape and size and are difficult to use to calculate PDD, BSF, or the output factor. The equivalent square can be calculated by the formula (refs 5, 6):

$$s = \frac{4A}{P}, \tag{1.29}$$

where s is the side of equivalent square, A is the area of the rectangular field (for a rectangle of sides a and b = a×b), and P is the perimeter of the rectangle [for rectangle = 2(a+b)], which reduces the above formula to s = 2ab/a+b.

Tissue-Air Ratio (TAR)

The PDD was used as the primary dosimetric variable when treatment techniques were predominantly SSD.

- When newer machines began to be produced that could rotate around an isocenter, it became possible to treat patients using isocentric techniques. In an isocentric technique, the distance from the source to the center of the target volume is held constant while the distance from the source to the surface of the patient changes for each beam orientation.

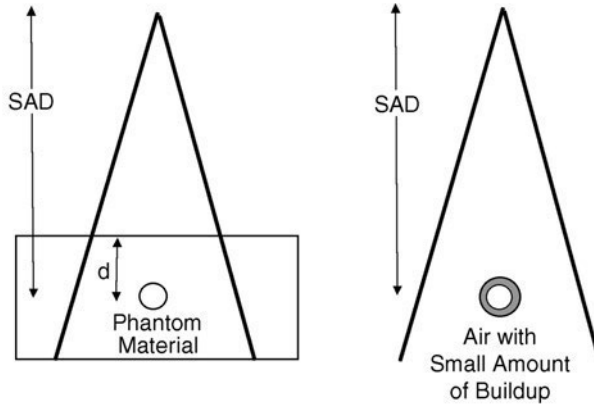


Figure 1–19. Tissue-air ratio (TAR) measurement: (a) dose measured at depth d in phantom; (b) dose measured in free space at the same point. (SAD = Source-to-Axis distance.)

- A new quantity was then defined to address isocentric treatment calculations, called the *tissue-air ratio, or TAR* (see Figure 1–19). The TAR is defined as:

$$\text{TAR} = \frac{\text{Dose at depth d in phantom}}{\text{Dose in free space at the same point}}. \qquad (1.30)$$

Note that the TAR varies like the PDD with respect to the beam energy, depth, and the field size; it is independent of SSD.

Scatter-Air Ratio (SAR)

When dealing with photon beams, a two-component model can be used to describe the radiation beam. In this model, the absorbed dose to any point in the patient is the sum of two components:

$$\text{Total dose} = \text{Primary dose} + \text{Scatter dose}. \qquad (1.31)$$

- **Primary Dose Component:** The primary component consists of all photons that come from the head of the machine that contribute dose to the point of interest.
- **Scatter Dose Component:** The scatter component consists of all photons that are scattered from the machine and the patient that contribute dose to the point of interest.

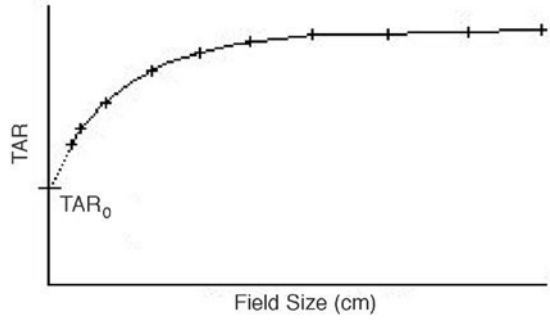


Figure 1–20. Plot of TAR as a function of field size.

If we plot the TAR as derived above, as a function of field size, we can extrapolate from the smallest fields measurable to a hypothetical zero area field (see Figure 1–20). This zero area field is an abstraction and is said to represent the primary dose (a field that contains no scatter).

Now the scatter component can be found by subtracting the zero-area TAR from all other TARs at that depth (a field that contains only scatter), given by:

$$SAR = TAR - TAR_0. \tag{1.32}$$

This quantity is called the *scatter-air ratio (SAR)* and is utilized in the calculation of dose in irregular fields.

Tissue-Phantom Ratio (TPR)

The definition of TAR is not easily measured above 3 MeV. Because of this limitation, a new dosimetric quantity was created for high-energy photon beams. The *tissue-phantom ratio (TPR)* is defined to be:

$$TPR = \frac{\text{Dose at depth d in phantom}}{\text{Dose at a specified reference depth in phantom}}, \tag{1.33}$$

where the specified reference depth is chosen to be 5, 7, or 10 cm.

Tissue-Maximum Ratio (TMR)

If the specified reference depth of the above TPR is redefined to be at the depth of maximum dose, d_{max}, then we have a special case of a TPR called the *tissue-maximum ratio (TMR)* (see Figure 1–21):

$$TMR = \frac{\text{Dose at depth d in phantom}}{\text{Dose at a depth } d_{max} \text{ in phantom}}. \tag{1.34}$$

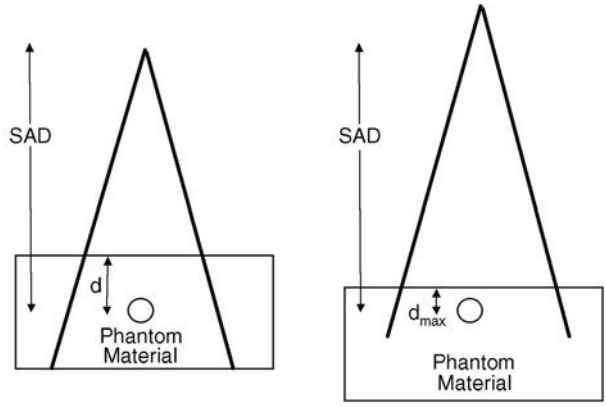


Figure 1–21. Tissue-maximum ratio (TMR) measurement: (a) dose measured at depth in phantom; (b) dose measured at d_{max} in phantom.

This quantity is the most frequently used for calculations of beam time and dose for isocentric techniques used currently in radiotherapy.

Calibrated Output

A treatment machine is calibrated before the start of clinical use. This calibration is done as per the American Association of Physicists in Medicine Task Group 51 (AAPM TG-51) protocol (ref 7). The protocol specifies that for high-energy photons, the calibration is done for a reference field size (10×10 cm^2), at a reference depth of 10 cm in water, and at a reference treatment distance from the source (SAD or SSD = 100 cm). An ionization chamber is used to measure the raw ionization for a set number of MUs at the reference depth in the water phantom. The raw ionization reading is fully corrected to ionization reading M as below:

$$M = P_{ion} \times P_{tp} \times P_{elec} \times P_{pol} \times M_{raw}, \tag{1.35}$$

where P_{ion} corrects for incomplete ion collection efficiency, P_{tp} corrects for temperature and pressure at the time of measurement, P_{elec} corrects for the response of the electrometer, P_{pol} corrects for any polarity effects, and M_{raw} is the raw ion chamber reading in coulombs (C).

The fully corrected ionization reading M is then used to calculate the dose D_w to water at depth as below:

$$D_w = M \times k_q \times N_{d,w}. \tag{1.36}$$

where M is the fully corrected ion chamber reading, k_q is the beam quality conversion factor which converts the calibration factor from a Co-60 beam to that for a beam quality q, and $N_{d,w}$ is the absorbed dose to water calibration factor for the ion chamber, which is given by an Accredited Dosimetry Calibration Laboratory (ADCL).

The dose at depth D_w is then divided by the PDD (for SSD calibration) or TMR (for SAD calibration) at 10 cm depth to determine the dose at d_{max}. This calibration factor that is determined serves as the foundation for all of the treatments on the treatment machine. This number is called the *calibrated output, which is set at 1 cGy/MU* and is used to calculate the necessary MUs needed to deliver the tumor dose prescription for a patient.

Collimator Scatter Factor and Phantom Scatter Factor

The effects of field size on the output of a treatment machine for different field sizes can be determined with the collimator factors, defined as:

- **Collimator Scatter Factor (S_c):** The collimator scatter factor takes into account radiation dose that originates only from scatter of the collimator and head of the machine.
- **Phantom Scatter Factor (S_p):** The radiation dose that originates only from scatter from the phantom/patient.
- **Collimator-Phantom Scatter Factor ($S_{c,p}$):** Since it is difficult to measure directly the dose from the phantom only, the collimator-phantom scatter factor is defined as any scattered radiation originating from the collimator, head, or patient.

Beam Modifier Factors

The effect of any beam modifiers, such as Lucite™ trays and wedges, must also be taken into account, since these devices cause a decrease in the output of the machine. The factors that take into account this attenuation of the beam are called *tray factors (TF)* and *wedge factors (WF)* defined by:

$$\begin{aligned} TF &= \frac{\text{Dose with tray in field}}{\text{Dose without tray in field}} \\ WF &= \frac{\text{Dose with wedge in field}}{\text{Dose without wedge in field}}. \end{aligned} \tag{1.37}$$

Monitor Unit Calculation

Now that we have all the basic dosimetric parameters defined, we can calculate the MUs necessary to deliver a prescribed field dose for a source-to-axis (SAD) radiation treatment. Monitor units are computed so that proper radiation dose can be delivered for each beam during the treatment. For an isocentric setup the MU needed to deliver the prescribed tumor dose per field of TD_{iso} are:

$$MU = \frac{TD_{iso}}{D_{ref} \cdot TMR(r_{bl}, d) \cdot S_c(r_{op}) \cdot S_p(r_{bl}) \cdot TF \cdot WF \cdot \left(\frac{SCD}{SAD}\right)^2}, \quad (1.38)$$

where TD_{iso} is the prescribed tumor dose, D_{ref} is the calibrated output 1 cGy/MU, TMR is tissue-maximum ratio for the blocked field, S_c is the collimator scatter factor for the open field, S_p is the phantom scatter factor for the blocked field, TF is the tray factor, WF is the wedge factor, SCD is the distance from the source to the calibration point, and SAD is the distance from the source to the isocenter (axis).

Electron Beam Dosimetry

Electrons behave differently from x-rays. Because of their unique interaction with matter they demonstrate a rapid dose fall-off, a finite range, and x-ray contamination. Unlike high-energy x-rays, electrons have less skin sparing and have a high surface dose. As such, in addition to the depth of maximum dose denoted as R_{100}, electron beams are described by other unique parameters such as R_{90}, R_p, and x-ray component as seen below in Table 1–6.

- **R_{90}** is the depth of 90% dose level and is the recommended therapeutic range of the electron beam depth dose.
- **R_p** is the practical range of the electron beam; that is, the maximum distance an electron beam can penetrate. This is the depth where the fall-off tangent of the electron beam depth dose curve intersects with the bremsstrahlung component line.
- Due to the interaction of the electron beam with the accelerator head components, *bremsstrahlung x-rays* are produced. This is known as bremsstrahlung contamination of the electron beam, which produces a tail in the electron beam depth dose. The bremsstrahlung component of the electron beam is calculated by extrapolating the bremsstrahlung tail to the R_p.

Table 1–6. Electron Beam Dose of Maximum Depth, R_{90}, and R_p Varies with Beam Energy*

Electron Beam Central Axis Depth Dose Data, SSD = 100 cm			
Nominal energy (MeV)	**R_{100} (d_{max}) (cm)**	**R_{90} (cm)**	**R_p (cm)**
6	1.3	1.7	2.88
9	1.8	2.6	4.32
12	2.3	3.5	5.85
15	2.5	4.4	7.26
18	2.5	5.2	9.25

*Data from ref 3, Jani SK, *Handbook of Dosimetry Data for Radiotherapy,* Table 2.A1, p. 76, 1993.

To make the electron beam clinically useful, the beam requires collimation with electron cones on account of the collisions that electrons undergo with air. In addition, electron custom blocks are placed at the electron cone ends to shape the electron beam to conform to the treatment volume. Note that this blocking affects the electron beam output and the beam PDD. Electron custom block output is routinely measured before clinical use. Nevertheless, with proper attention and beam modifiers such as cutout and skin boluses, electron beams are very useful for superficial targets such as tumor beds or skin cancers.

THE PHYSICS OF RADIATION TREATMENT PLANNING AND DELIVERY

Radiation treatments can be done in two different ways:

- External beam radiation therapy, in which radiation is given from a linear accelerator.
- Brachytherapy (internal radiation), in which radiation is given by implanting a radioactive source into the patient.

External Beam Radiation

Modern external beam radiation therapy is performed via a linear accelerator (see Figure 1–22), which accelerates electrons to very high energies. The electrons strike a tungsten target to produce bremsstrahlung x-rays, which are then used to treat patients. Alternatively, high-energy

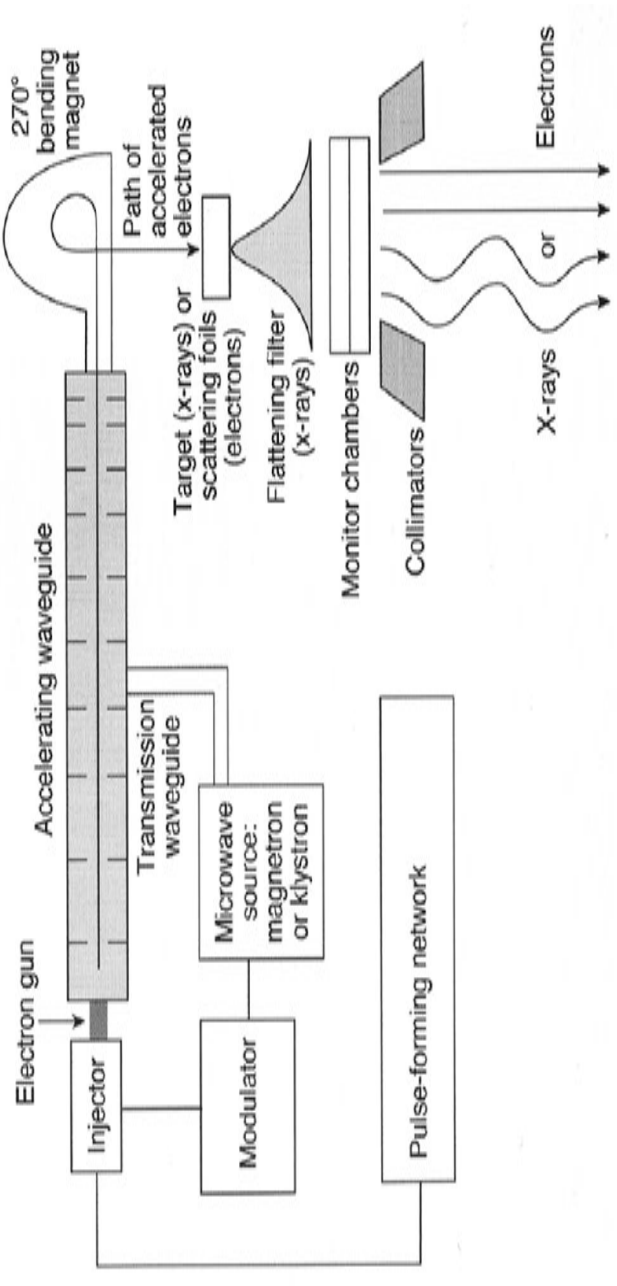


Figure I–22. Schematic diagram of a linear accelerator. (Reprinted from ref 8, Gunderson L and Tepper J. *Clinical Radiation Oncology*, 1st edition, Fig. 3-5, p. 68, © 2000, with permission from Elsevier.

electrons can also be used for the patient's treatment. Better skin sparing and better penetration of the tissue make a linear accelerator an optimum tool for radiation treatments.

- Modern linear accelerators can produce both low- and high-energy photons, such as 6 and 18 MV photons, and ranges of electrons from 4 to 20 MeV.
- In addition, modern linear accelerators are now equipped with dual asymmetrical jaws, a multileaf collimator (MLC), and electronic portal imaging devices (EPIDs). The dual asymmetric jaws can be used to generate dynamic wedges with nominal wedge angles ranging from 10° to 60°, as opposed to fixed physical wedges. The standard MLC consists of individual collimator leaves of 0.5 cm to 1 cm at the isocenter, which are individually controlled by a computer.
- Although initially designed to replace Cerrobend® blocks, the MLC is now used to modulate the beam intensity to perform intensity-modulated radiation therapy (IMRT).
- In addition, with the use of new imaging technologies EPIDs are now in routine clinical use. Not only automating the portal imaging to verify patient positioning, EPIDs are also essential to perform image-guided radiation therapy (IGRT) for more accurate delivery of the radiation.

Internal Radiation

Internal radiation is also called *brachytherapy*. Radioactive sources, such as ^{125}I seeds, ^{137}Cs capsules, or ^{192}Ir ribbons are placed inside the patient in the treatment areas. These radioactive sources then deliver high-dose radiation to the tumor with a rapid dose fall-off, sparing the surrounding normal critical structures from the high dose of radiation. Brachytherapy can be interstitial or intracavitary, and can be temporary or permanent.

Radiation Treatment Planning

After a patient is counseled and a decision has been made for external beam radiation treatment, a planning process is set into motion. For a conformal radiation therapy technique the patient is CT simulated, which involves the following process:

- *The patient is set in the treatment position:* The exact treatment position for the patient is instructed by the treating physician. This may include specific positioning of the patient such as supine or prone, position of the head, arms, legs, etc. A treatment aid such

as a breast board or belly board may be required, depending upon the site of the treatment. To reproduce the treatment setup, it is important that the patient be in a comfortable position.

- *Immobilization of the patient:* Immobilization devices ensure the setup reproducibility on a daily basis. These devices include thermoplastic face masks for brain and head and neck patients, wing boards for lung patients, and body molds for breast and prostate patients. Often, immobilization devices are custom-made, thus offering reproducibility and rigidity.
- *CT scanning:* The superior and inferior limits of the CT scan are instructed by the treating physician. In addition, he or she may instruct the use of contrast, either given orally and/or intravenously for greater precision of tumor localization. All surgical scars are usually marked by radiopaque wires for identification on CT images. Depending upon the no-shift or shift methods, reference marks may be placed on the patient prior to CT scans.
- *Transfer of CT dataset to the treatment planning station:* Once the CT scan is complete, the CT dataset is then transferred to the planning station, where the data are prepared by the dosimetrist for a virtual simulation. This involves the orientation of the patient/treatment coordinate system without the presence of the patient. The three-dimensional dataset (3D CT) enables the radiation oncologist to make treatment planning decisions that he would have been unable to do during conventional simulation. In addition, at the treatment planning station any secondary imaging dataset such as diagnostic CT, magnetic resonance imaging (MRI), or positron emisson tomography (PET) can be registered and fused with the primary CT dataset for greater localization of the target volume.

At this point the CT simulation process has ended and the planning process has started, involving the following steps:

- *Contouring structures:* Once the dataset is prepared by the dosimetrist at the planning station, the treating physician can then start contouring the critical structure and the target volumes. More recently, physicians use the International Commission on Radiation Units and Measurements (ICRU) reports 50 and 62 (refs 9, 10) definitions for different target volumes (see Figure 1–23). Briefly, gross target volume (GTV) is the gross disease seen clinically or in imaging studies, clinical target volume (CTV) is the clinically suspected extension of the tumor beyond the gross disease, internal target volume (ITV) is clinical target volume and a margin to account for the internal movement of the CTV, and planning target volume (PTV) is the margin needed for the uncertainty of the setup error for the patient. In addition to target volumes, critical structures, organ at risk (OAR) volume has also been defined by the ICRU. Once the critical structures are outlined

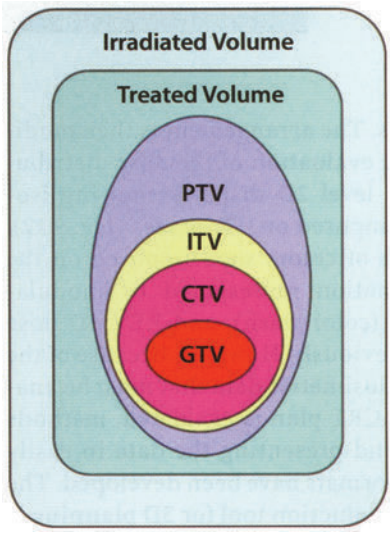


Figure 1–23. Different target volumes as defined by the International Commission on Radiation Units and Measurements (ICRU). (Reprinted from ref 11, Levitt SH, Purdy JA, Perez, CA, Vijayakumar S. *Technical Basis of Radiation Therapy: Practical Clinical Applications,* 4th edition, Fig. 9.13, p. 188, © 2006 with kind permission of Springer Science and Business Media.)

by the radiation oncologist, a margin is then added to the OAR for the uncertainty of the setup error for the patient. This is called the planning organ at risk volume (PRV).

- *Beam placement and design of the treatment fields:* Once all the critical structures and the target volumes are contoured, the dosimetrist then initiates the radiation beam placement as per the physician instruction. Standard setup treatments such as four-field box pelvis requires only four beams coming from AP/PA and RT/LT lateral beams; other treatments such as nine-field IMRT may require complex beam arrangements. Modern treatment planning systems use MLCs for normal tissue blocking and use dynamic wedges as opposed to Cerrobend blocks and physical wedges of the past. Depending on the target volumes and the prescription, appropriate treatment fields are then designed by the dosimetrist. These fields should be verified by the physician prior to initiating the dose calculation.
- *Dose calculation and isodose generation:* The next step in the process of the radiation treatment planning is to generate the 3-D dose distribution for the patient. Highly sophisticated computer software is used to derive an optimized treatment plan for each patient. In general, isodose distribution in all three axes and a

dose-volume histogram (DVH) is generated for the physician to review and approve the plan.

- *Treatment plan verification and quality assurance (QA):* The final step of the treatment planning is the verification of the treatment plan that has been designed for an individual patient. For a simple setup this may involve taking only setup verification films prior to initiating the radiation treatment to verify the bony landmarks; for more complex IMRT plans, detailed measurements of the fluence map are needed.

Once the planning is complete and approved by the radiation oncologist, necessary QA is then done by the medical physicist. The treatment planning information is then transferred to the record and verify (R&V) system for delivery of the radiation therapy by the linear accelerator.

External Beam Radiation Therapy

Radiation treatment planning has come a long way in the last 20 years. Powerful computer hardware, coupled with sophisticated software, now enables radiation treatment planning to provide state-of-the-art image processing, precise target localization, and innovative 3-D dose computation. Commercially available radiation dose calculation algorithms now range from pencil-beam algorithms to the most precise Monte Carlo algorithm. The following sections describe radiation treatment modalities.

3-D Conformal Radiation Therapy (3D-CRT)

Computer-generated reconstruction of the ICRU-defined target volumes and surrounding OARs is done from direct CT/MRI data.

- The dosimetrist/radiation oncologist team then uses a beam's-eye view (BEV) (the relationship of the target and normal tissue as seen by the radiation beam) to arrange radiation portals in preparation for non-coplanar/coplanar radiation therapy to treat the tumor. 3D-CRT is a forward planning technique. This is a manual trial and error and iterative process performed by the dosimetrist, who continuously evaluates treatment plans by altering beam parameters (such as beam directions, beam weights, beam modifiers, etc.) until a satisfactory plan is achieved. 3D-CRT dose calculation is more accurate as the dose computation algorithm uses 3-D datasets from the CT scan, providing comprehensive patient anatomy; and 3-D dose corrections are done which take into account tissue heterogeneities. To perform a 3D-CRT plan evaluation, isodose distribution in three-dimensional planes is most useful as it aids the radiation oncologist in determining the spatial relationship of the radiation dose to the target and the OARs. However, since the biologic effect of the radiation is also a function of the volume of the organ irradiated, a DVH becomes

a critical tool for treatment plan evaluation. It is the cumulative DVH, which plots the volume of the targets and critical structures as a function of dose, that is most useful to the radiation oncologist. An optimized treatment plan should include at least 95% of the PTV within the 100% of the prescribed isodose line, while maintaining all the critical structure constraints set forth by the physician.

Intensity Modulated Radiation Therapy (IMRT)

IMRT is a newer technology in radiation oncology that delivers highly conformal radiation to the tumor while improving normal tissue sparing. It combines two advanced concepts to deliver 3-D conformal radiation:

(1) Inverse treatment planning, utilizing techniques such as simulated annealing or gradient optimization. Most inverse optimization problems in radiation therapy are simple and straightforward and can be solved by a gradient optimization technique, which is also much faster and is used by many treatment planning systems.

(2) Computer-controlled intensity modulation of the radiation beam. In IMRT radiation is broken into many "beamlets"; the intensity of each beamlet can be adjusted individually and is achieved by the treatment planning program.

- Once all the target volumes and OARs are outlined, five to nine equispaced coplanar beams are placed around the patient, and a low energy of 6 MV photons is usually selected as the beam energy. The large number of beam orientations may nullify any requirement for beam angle optimization for the IMRT planning.
- The radiation oncologist then mathematically quantifies the desired dose distribution within the patient by defining some objective functions such as assigning minimum and maximum doses acceptable for the treating targets and minimum and maximum limiting doses for the OARs. For example, a prescription plan for prostate cancer IMRT treatment might be: prostate PTV dose 180 cGy/fx to 7920 cGy, the rectal constraint V70 $<$ 25% and V40 $<$ 35%, bladder constraint V75 $<$ 25%, and V40 $<$ 40%, femoral head constraint is V50 $<$ 5% and V45 $<$ 25% and maximum penile bulb dose $<$52 Gy. This type of plan optimization is known as dose-volume–based optimization for IMRT.
- Once a satisfactory IMRT plan is achieved, the plan evaluation is done by reviewing isodose distribution in three-dimensional planes of axial, coronal, and sagittal and by reviewing the cumulative DVH. However, very careful attention must be given to IMRT plan evaluation as unexpected high doses can be deposited in unintended locations within the patient. An optimized treatment plan should include at least 95% of the PTV within the 100% of the prescribed isodose line, while maintaining all the critical structure constraints set forth by the physician. The minimum and the maximum dose constraints for the PTV should be within 5% of the prescription dose. This helps to achieve a more homogeneous dose distribution within the target volume.

- Once the IMRT plan is approved by the radiation oncologist, the non-uniform beam fluence map is then converted into the MLC sequence for the accelerator. The beams are delivered by computer-controlled opening and closing of the MLC leaves.
- There are two main types of IMRT delivery systems: (1) static multileaf collimation (sMLC), commonly known as the step-and-shoot technique and (2) dynamic multileaf collimation (dMLC), commonly known as the sliding window technique. In the sMLC technique, while the gantry is in one angle, the non-uniform fluence is delivered by using a sequence of fixed, multiple, small subfields, usually between 20 and 100. The radiation stops while the MLC pattern changes from one subfield to the other at each gantry angle. In the dMLC technique the MLC leaves are continuously moving while the radiation beam is on. It is the slow and fast variable speeds of the MLC leaves that cause the high and low intensity of the non-uniform fluence of the beam during the treatment.

The potential advantages of IMRT are to create multiple targets and multiple critical avoidance structures and to use new accelerated fractionation schemes, such as simultaneous integrated boost (SIB) treatments. IMRT is the most important advance in radiotherapy since the invention of the linear accelerator.

Image-Guided Radiation Therapy (IGRT)

Setup errors and organ motion contribute significant uncertainties during radiation therapy delivery. To overcome this problem, modern linear accelerators are equipped with portal vision and/or an on-board imager. This has allowed the radiation oncologist to move from two-dimensional (2-D) portal films for setup verification to 3-D volumetric soft tissue imaging of the target for treatment verification.

- X-ray images can be taken prior to daily radiation treatments to make necessary positional adjustments to complete the radiation treatment with millimeter accuracy. These are called *online corrections*.
- For example, for image guided prostate cancer treatments non-radioactive fiducial gold marker seeds are implanted into the prostate. Portal images/cone beam CTs are taken prior to daily treatment to localize these marker seeds. Table shifts are then made in the *x*-, *y*-, *z*-directions, based on the shifted marker seed location compared to the initial CT/simulation digitally reconstructed radiographs (DRRs) every day.

The use of IMRT coupled with IGRT can now allow for tighter PTV margins, enabling the radiation oncologist to deliver a higher radiation dose to the tumor and improving sparing of normal critical structures safely.

BRACHYTHERAPY

In modern radiation oncology a large number of patients are treated with brachytherapy. Radioactive sources that are implanted into the tumor give a very high radiation dose to the tumor while giving a low dose to the surrounding tissue. This is because of the inverse square law and the attenuation of the radiation by the tissue. As the distance from the radioactive source increases, rapid dose fall-off in the surrounding normal tissue is observed. Brachytherapy can be done utilizing either an interstitial or an intracavitary technique.

Interstitial Brachytherapy

In this technique the radioactive sources are implanted directly into the tumorous tissue. Radioactive sources can be in the form of seeds, needles, or ribbons. In a temporary implant these radioactive sources can be removed from the tissue after a specific dose has been given; alternatively, the sources can be left in the implanted tissue as a permanent implant.

- One of the common uses of an interstitial implant is for treating prostate cancer. Prostate cancer implants can be permanent or temporary. The permanent implant is most commonly performed with either ^{125}I or ^{103}Pd radioactive seeds. Table 1–7 lists their dosimetric characteristics.
- One can see from the data in Table 1–7 that the radioactive sources used for permanent implants have short half-lives and low radiation energies, indicating that the majority of the radiation dose is given over a short period of time to a limited volume of tissue.
- On the other hand, the radioactive sources used for temporary implants have a longer half-life and high radiation energies. For temporary prostate cancer implants, the radioactive source ^{192}Ir is commonly used. ^{192}Ir has a half-value layer (HVL) of 3 cm of lead, average energy of 380 keV, and a half-life of 74.2 days, dose rate constant of 4.55 cGy mCi^{-1} h^{-1}.

The commercially available HDR units are proof that we have come a very long way from the Paterson-Parker or the Manchester system tables that were devised for dosimetric treatments in the early days of brachytherapy planning.

Table 1–7. Dosimetric Characteristics for ^{125}I and ^{103}Pd

	^{125}I	^{103}Pd
HVL tissue	2 cm (0.025 mm lead)	1.6 cm (0.008 mm lead)
Average energy	28 KeV	21 KeV
Half life	59.6 d	17 d
Dose rate constant	1.16 cGy mCi^{-1} h^{-1}	0.95 cGy mCi^{-1} h^{-1}
Dose rate	7 cGy/h	21 cGy/h
Primary dose	145 Gy	115 Gy
mCi	.31–.37 mCi/seed	1.4 mCi/seed
Boost dose	108 Gy	90 Gy
mCi	.25–.31 mCi/seed	1 mCi/seed

Intracavitary Brachytherapy

In this technique hollow stainless steel applicators hold the radioactive sources in the desired configuration for the brachytherapy treatments. This technique is most commonly used for cervical and uterine cancer treatments.

- Fletcher-Suit applicators, the most commonly used, have a tandem that is inserted into the uterine canal and colpostats that are inserted into the vaginal fornices, which are then secured with a locking mechanism. Once the applicators are stabilized with packing and verified with x-ray images, the applicators are then loaded with radioactive sources.
- Intracavitary low dose rate (LDR) brachytherapy is commonly done using ^{137}Cs radioactive sources. ^{137}Cs has an HVL of 6 mm of lead, average energy of 662 keV, and a half-life of 30 years, dose rate constant of 3.09 cGy mC^{-1} h^{-1}.
- Recently, a high activity 10 Ci source of ^{192}Ir with a dose rate of about 100 cGy/min, known as high dose rate (HDR), is in use for the temporary implants for prostate and other cancers. HDR implantation is often done via a remote afterloading technique. The HDR remote afterloading technique allows the brachytherapy treatment time to be short (in the magnitudes of minutes), spares personnel from radiation exposure, and with the aid of sophisticated treatment planning software to manipulate the source dwell times, adds flexibility to customize the brachytherapy treatments.

OTHER RADIATION THERAPY MODALITIES

Modern radiation oncology employs many other radiation therapy techniques for patient treatments.

- **Stereotactic radiosurgery (SRS)** is a technique that allows the radiation oncologist to focus beams of radiation precisely to treat tumors in a single-fraction treatment. Radiosurgery can be performed using x-rays from linear accelerators or gamma rays from Co-60 radioactive sources from a Gamma Knife® unit. Co-60 has an HVL of 11 mm of lead, average energies of 1.17 and 1.33 MeV, and a half-life of 5.26 years. SRS is mostly used for primary brain tumors and metastatic brain lesions. Similar to SRS, stereotactic body radiotherapy (SBRT), using a very small field over 3 to 5 fractions, has recently been used to treat lesions of the spinal cord, liver, adrenal gland, lung, and pancreas.
- **Particle radiation therapy** using proton and neutron beams is now in use to treat certain tumors that are very difficult to cure with photon radiation therapy. Proton beams deposit most of the radiation dose at a certain depth within the body (Bragg peak), which then better spares the underlying normal tissue; neutron beams have a greater relative biological effectiveness (RBE) on the tumor than a similar dose of photon radiation therapy.
- A **nonsealed radionuclide** such as ^{153}Sm is used intravenously to treat widespread, painful bony metastatic lesions. ^{153}Sm beta particles have a maximum range of 3 mm in soft tissue and 1.7 mm in bone, an average energy of 0.23 MeV, and a half-life of 1.9 days. On the other hand, radioimmunotherapy is a form of infusible treatment available for lymphoma cancer patients. Radioimmunotherapy uses cell-specific monoclonal antibodies which are radiolabeled with ^{111}In, ^{90}Y, or ^{131}I. These antibodies then bind to specific cancer cells (CD20 positive) and deliver radiation directly to the cancer cell.

A newer generation of external beam radiation therapy techniques is now becoming available. These includes Rapid Arc® radiation therapy and the CyberKnife®. The RapidArc™ radiation therapy is a newest approach to deliver image-guided–intensity-modulated radiation therapy (IG-IMRT). An On-Board Imager®, much like a CT scanner, is attached with this machine, which takes images of the exact location, shape, and size of the tumor, immediately before each radiation treatment is delivered. The images are reviewed in real time to make patient positional adjustment by moving the treatment table automatically to put the patient exactly where he needs to be, the same way every day within millimeter accuracy. The RapidArc machine then varies three parameters simultaneously to deliver precision radiation treatment: (1) the speed

with which the radiation machine rotates around the patient; (2) the dimensions of the beam shaping aperture, that molds the radiation beam to precisely fit the shape and size of the tumor; and (3) the rate at which the radiation dose is given to the patient. As such, this latest radiation technique can turn a 20-minute treatment time into a highly precise 90-second treatment time for some selected cancer patients. By shortening the treatment time, the effect of the patient's breathing and involuntary movement during the treatment can also be minimized, thus improving further tumor targeting accuracy.

CyberKnife uses a noninvasive image-guided localization system and robotic delivery of SRS to treat the patient. It is the robotic arm of the CyberKnife that gives this technique an unprecedented access to tumor sites with an ease that is very difficult to achieve with conventional radiation therapy technique. Most recently, this system is being tested to treat prostate cancer patients with four to five fractions of treatments in hypofractionated fashion, more like the HDR brachytherapy treatments for prostate cancer.

In the future, these exciting newer generations of radiation therapy will allow us to treat cancer patients with higher radiation doses, thus increasing the chance of cure, while minimizing side effects from the radiation treatments.

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