STRUCTURE OF MATTER

- Atomic dimensions: radius of atom ~10^{-10} m, radius of nucleus ~10^{-15} m.
- Classification of atoms
    - Isotopes—atoms with the same Z, different number of neutrons.
    - Isotones—atoms with the same number of neutrons, different Z.
    - Isobars—atoms with the same A, different Z.
    - Isomers—atoms with the same A and same Z but different nuclear energy states.
- Nuclear stability
    - Certain combinations of number of neutrons (n) and protons (p) in the nucleus show more stability than others.
    - The most stable nuclei contain even numbers of n and even numbers of p. Only exceptions are ${}^{2}_{1}H$, ${}^{6}_{3}Li$, ${}^{10}_{5}B$, and ${}^{14}_{7}N$.
    - The least stable nuclei contain odd numbers of n and odd numbers of p.
    - High n/p ratio gives rise to β^- decay and a low n/p ratio can result in electron capture and β^+ decay .
    - Beyond atomic number 82, all nuclides are radioactive.
- Number of electrons per gram = N_A. Z/A_W, where N_A is Avogadro's number, Z is atomic number, and A_W is the atomic weight.
- Elementary particles
    - There are 12 fundamental particles of matter: six quarks and six leptons. Correspondingly, there are six quarks and six leptons of antimatter. All these particles are called fermions. In addition, there are 13 messenger particles, called bosons, that mediate the four forces of nature.
    - Fermions have a noninteger spin; bosons have an integer spin.
    - The Higgs field permeates all space and is responsible for giving mass properties to matter. The messenger particle for the Higgs field is the Higgs boson

- Forces of nature
    - There are four forces of nature. In order of their strengths, they are strong nuclear, electromagnetic, weak nuclear, and gravitational.
    - All forces of nature are mediated by specific messenger particles, the bosons.
- Wavelength (λ), frequency (n), and velocity (c) of electromagnetic waves are related by c = nl.
- The quantum model relates energy of a photon with its frequency of oscillation by E = hn, where h is the Planck's constant.

NUCLEAR TRANSFORMATIONS

The term half-life ($T_{1/2}$) of a radioactive substance is defined as the time required for either the activity or the number of radioactive atoms to decay to half the initial value. The mean or average life is the average lifetime for the decay of radioactive atoms. Although, in theory, it will take an infinite amount of time for all the atoms to decay, the concept of average life (T_a) can be understood in terms of an imaginary source that decays at a constant rate equal to the initial activity and produces the same total number of disintegrations as the given source decaying exponentially from time $t = 0$ to $t = \infty$.

$$T_a = 1.44\ T_{1/2}$$

The activity per unit mass of a radionuclide is termed the specific activity.

The specific activity of radium is slightly less than 1 Ci/g, although the curie was originally defined as the decay rate of 1 g of radium (0.975 Ci). The reason for this discrepancy, as mentioned previously, is the current revision of the actual decay rate of radium without modification of the original definition of the curie.

Radioactive series:

All naturally occurring radioactive elements have been grouped into three series: the uranium series, the actinium series, and the thorium series.

The uranium series originates with ^{238}U having a half-life of 4.51×10^9 years and goes through a series of transformations involving the emission of α and β particles. γ rays are also produced as a result of some of these transformations. Ends with lead 206.

The actinium series starts from ^{235}U with a half-life of 7.13×10^8 years and ends with lead 207.

The thorium series begins with ^{232}Th with a half-life of 1.39×10^{10} years and ends with lead 208.

Radioactive equilibrium:

Many radioactive nuclides undergo successive transformations in which the original nuclide, called the parent, gives rise to a radioactive product nuclide, called the daughter. If the half-life of the parent is longer than that of the daughter, then after a certain time, a condition of equilibrium will be achieved; that is, the ratio of daughter activity to parent activity will become constant. In addition, the apparent decay rate of the daughter nuclide is then governed by the half-life or disintegration rate of the parent.

Two kinds of radioactive equilibria have been defined, depending on the half-lives of the parent and the daughter nuclides.

If the half-life of the parent is not much longer than that of the daughter, then the type of equilibrium established is called the **transient** equilibrium. Example : parent ^{99}Mo ($T_{1/2}$ = 67 hours) and the daughter ^{99m}Tc ($T_{1/2}$ = 6 hours).

Iif the half-life of the parent is much longer than that of the daughter, then it can give rise to what is known as the **secular** equilibrium. Example: parent ^{226}Ra ($T_{1/2}$ = 1,622 years) and daughter ^{222}Rn ($T_{1/2}$ = 3.8 days).

Modes of radioactive decay:

<u>Alpha particle decay:</u>

Radioactive nuclides with very high atomic numbers (>82) decay most frequently with the emission of an α particle. It appears that as the number of protons in the nucleus increases beyond 82, the Coulomb forces of repulsion between the protons become large enough to overcome the nuclear forces that bind the nucleons together. Thus, the unstable nucleus emits a particle composed of two protons and two neutrons. This particle, which is in fact a helium nucleus, is called the α particle.

A typical example of α decay is the transformation of radium to radon:

$$^{226}_{88}\mathrm{Ra} \xrightarrow[1{,}622\ \text{years}]{T_{1/2}} {}^{222}_{86}\mathrm{Rn} + {}^{4}_{2}\mathrm{He} + 4.87\ \mathrm{MeV}$$

<u>Negatron decay:</u>

The radionuclides with an excessive number of neutrons or a high neutron-to-proton (n/p) ratio lie above the region of stability. These nuclei tend to reduce the n/p ratio to achieve stability. This is accomplished by emitting a negative electron.

$$^{32}_{15}\mathrm{P} \xrightarrow[14.3\ \text{days}]{T_{1/2}} {}^{32}_{16}\mathrm{S} + {}^{0}_{-1}\beta + \tilde{\nu} + 1.7\ \mathrm{MeV}$$

<u>Positron decay:</u>

Positron-emitting nuclides have a deficit of neutrons, and their n/p ratios are lower than those of the stable nuclei of the same atomic number or neutron number. For these nuclides to achieve stability, the decay mode must result in an increase of the n/p ratio. One possible mode is the β decay involving the emission of a positive electron or positron.

$$^{22}_{11}\mathrm{Na} \xrightarrow[2.60\ \text{days}]{T_{1/2}} {}^{22}_{10}\mathrm{Ne} + {}^{0}_{+1}\beta + \nu + 1.82\ \mathrm{MeV}$$

<u>Electron capture:</u>

The electron capture is a phenomenon in which one of the orbital electrons is captured by the nucleus, thus transforming a proton into a neutron.

Nuclear fission and fusion:

- Nuclear fission is a process of splitting high Z nucleus into two lower Z nuclei. The process results in the release of a large amount of energy. Example: fission of ^{235}U nucleus by bombarding it with thermal neutrons (i.e., neutrons of energy <0.025 eV). A chain reaction is possible with a critical mass of fissionable material.
- Nuclear fusion is the reverse of nuclear fission—lighter nuclei are fused together into heavier ones. Again, a large amount of energy is released in the process. Fusion of hydrogen nuclei into helium nuclei is the source of our sun's energy.

PRODUCTION OF X RAYS

At electron energies below about 100 keV, x-rays are emitted more or less equally in all directions. As the kinetic energy of the electrons increases, the direction of x-ray emission becomes increasingly forward. Therefore, transmission-type targets are used in megavoltage x-ray tubes (accelerators) in which the electrons bombard the target from one side and the x-ray beam is obtained on the other side.
The probability of bremsstrahlung production varies with the Z^2 of the target material. However, the efficiency of x-ray production depends on the first power of atomic number and the voltage applied to the tube. The term efficiency is defined as the ratio of output energy emitted as x-rays to the input energy deposited by electrons.

The exposure rate is plotted as a function of the tube current. There is a linear relationship between exposure rate and tube current. As the current or milliamperage is doubled, the output is also doubled.

The increase in the x-ray output with increase in voltage is much greater than that given by a linear relationship. Although the actual shape of the curve depends on the filtration, the output of an x-ray machine varies approximately as a square of kilovoltage.

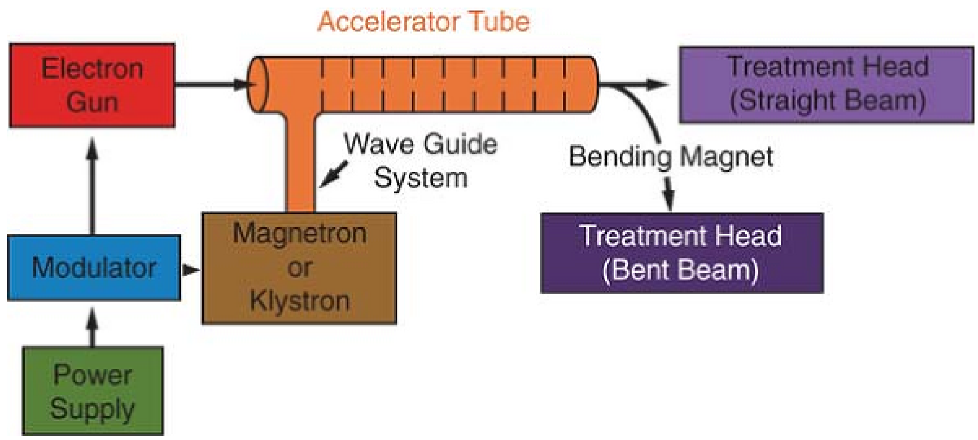
- X-rays are produced by two different mechanisms: bremsstrahlung and characteristic x-ray emission. Useful x-ray beams in imaging and therapy are all bremsstrahlung.
- Bremsstrahlung x-rays have a spectrum of energies. The maximum energy is numerically equal to the peak voltage. Average energy is about one third of the maximum energy.
- Characteristic x-rays have discrete energies, corresponding to the energy level difference between shells involved in the electron transition.

CLINICAL RADIATION GENERATOR

X-ray therapy in the kilovoltage range:

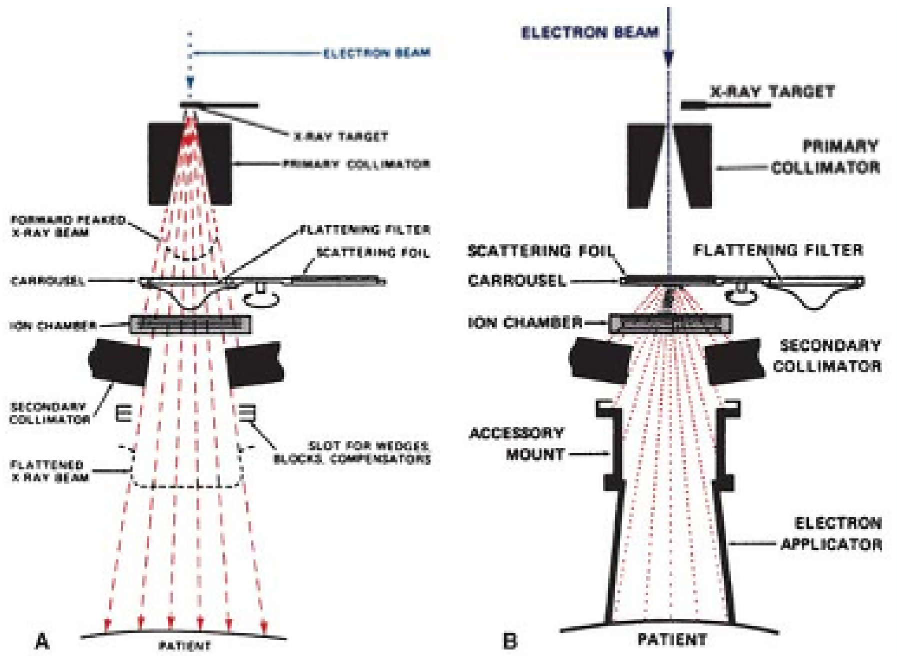
(1) Grenz ray therapy: <20kV, not used clinically
(2) Contact X rays: 40-50kV, used for endocavitary radiotherapy for early rectal cancers
(3) Superficial therapy:50-150kV, used for skin cancers
(4) Orthovoltage/ deep X rays: 150-500kV
(5) Supravoltage X rays: 500-1000kV

Working of a Linear Accelerator:



Power supply=DC
Modulator=pulse forming network and a switch tube (hydrogen thyratron)
Accelerator tube=accelerator wave-guide, evacuated copper tube lined by copper discs
Electrons of initial energy 50keV gain energy by interacting with the microwaves. When they strike the target, they have a width of 3mm. Bending magnets are required in high-energy LAs (>6MV)
Magnetron=microwave generator, present in lower-energy LAs (upto 6MV). Microwave frequency around 3000 MHz.
Klystron=microwave amplifier, present in higher-energy LA
Target=Tungsten. Brehmsstrahlung X rays are produced
Flattening filter=Lead, to make the X ray beam uniform in intensity across its expanse. It is thicker in the middle and thinner at the edges.
Scattering foil=Lead, to spread the electron beam

The treatment head onsists of a thick shell of high-density shielding material such as lead, tungsten, or lead-tungsten alloy. It contains an x-ray target, scattering foil, flattening filter, ion chamber, fixed and movable collimator, and light localizer system. Focal spot (virtual source) is 2-3mm in diameter.



<u>X ray beam</u>: Target$\rightarrow$Primary Collimator$\rightarrow$Flattening Filter$\rightarrow$Ion Chambers$\rightarrow$Secondary collimator[$\rightarrow$Tertiary collimator (MLC)$\rightarrow$Wedge/Block/Compensator]
<u>Electron beam</u>: Primary collimator$\rightarrow$Scattering foil$\rightarrow$Ion chamber$\rightarrow$Secondary collimator$\rightarrow$Eelctron applicator

The point of intersection of the collimator axis and the axis of rotation of the gantry is known as the isocenter.

Cobalt source:

The ^{60}Co source is produced by irradiating ordinary stable ^{59}Co with neutrons in a reactor. The nuclear reaction can be represented by ^{59}Co(n,γ) ^{60}Co.

The ^{60}Co source, usually in the form of a solid cylinder, discs, or pallets, is contained inside a stainless steel capsule and sealed by welding. This capsule is placed into another steel capsule, which is again sealed by welding. The double-welded seal is necessary to prevent any leakage of the radioactive material.

The ^{60}Co source decays to ^{60}Ni with the emission of β particles (E_{max} = 0.32 MeV) and two photons per disintegration of energies 1.17 and 1.33 MeV . These γ rays constitute the useful treatment beam. The β particles are absorbed in the cobalt metal and the stainless steel capsules, resulting in the emission of bremsstrahlung x-rays and a small amount of characteristic x-rays. However, these x-rays of average energy around 0.1

MeV do not contribute appreciably to the dose in the patient because they are strongly attenuated in the material of the source and the capsule. The other “contaminants” to the treatment beam are the lower-energy γ rays produced by the interaction of the primary γ radiation with the source itself, the surrounding capsule, the source housing, and the collimator system. The scattered components of the beam contribute significantly (~10%) to the total intensity of the beam (9). All these secondary interactions thus, to some extent, result in heterogeneity of the beam. In addition, electrons are also produced by these interactions and constitute what is usually referred to as the electron contamination of the photon beam.

Source diameter=1-2 cm

Penumbra:

Penumbra, in a general sense, means the region, at the edge of a radiation beam, over which the dose rate changes rapidly as a function of distance from the beam axis.

(1) The transmission penumbra is the region irradiated by photons that are transmitted through the edge of the collimator block. The penumbra width increases with an increase in source diameter, SSD, and depth but decreases with an increase in SDD.

(2) The geometric penumbra is due to the finite size of the source. It is independant of the field size. Its width is proportional to source diameter. It increases with increase in SSD and depth and decreases with increase in SDD.

(3) At a depth in the patient the dose variation at the field border is a function of not only geometric and transmission penumbras, but also the scattered radiation produced in the patient. Thus, dosimetrically, the term physical penumbra width has been defined as the lateral distance between two specified isodose curves (20%-90%) at a specified depth (10cm).

INTERACTIONS OF RADIATION WITH MATTER

- The fluence (Φ) of photons is the quotient dN by da, where dN is the number of photons that enter an imaginary sphere of cross-sectional area da.
- Fluence rate or flux density (φ) is the fluence per unit time.
- Energy fluence (ψ) is the quotient of dE_{fl} by da, where dE_{fl} is the sum of the energies of all the photons that enter a sphere of cross-sectional area da.
- Energy fluence rate, energy flux density, or intensity (c) is the energy fluence per unit time:

The reduction in the number of photons (dN) is proportional to the number of incident photons (N) and to the thickness of the absorber (dx).

$$dN = -\mu N dx$$

$$\frac{dI}{I} = -\mu dx$$

If thickness x is expressed as a length, then μ is called the linear attenuation coefficient. This coefficient depends on the energy of the photons and the nature of the material. Since the attenuation produced by a thickness x depends on the number of electrons presented in that thickness, μ depends on the density of the material. Thus, by dividing μ by density r, the resulting coefficient (μ/r) will be independent of density; μ/r is known as the mass attenuation coefficient. This is a more fundamental coefficient than the linear coefficient, since the density has been factored out and its dependence on the nature of the material does not involve density but rather the atomic composition. The mass attenuation coefficient has units of cm^2/g.

Half-value layer (HVL) defined as the thickness of an absorber required to attenuate the intensity of the beam to half its original value.

$$\mathrm{HVL} = \frac{0.693}{\mu}$$

As the filter thickness increases, the average energy of the transmitted beam increases or the beam becomes increasingly harder. Thus, by increasing the filtration in such an x-ray beam, one increases the penetrating power or the half-value layer of the beam.

At high photon energy, the Compton effect causes a large amount of energy absorption compared with the Compton interactions involving low-energy photons.

Thus, if the energy of the incident photon is high ($\alpha >> 1$), we have the following important generalizations:

- the radiation scattered at right angles is independent of incident energy and has a maximum value of 0.511 MeV;
- the radiation scattered backwards is independent of incident energy and has a maximum value of 0.255 MeV

Compton effect is an interaction between a photon and a free electron. Practically, this means that the energy of the incident photon must be large compared with the electron-binding energy. This is in contrast to the photoelectric effect, which becomes most probable when the energy of the incident photon is equal to or slightly greater than the binding energy of the electron.

The pair production process is an example of an event in which energy is converted into mass, as predicted by Einstein's equation $E = mc^2$. The reverse process, namely the conversion of mass into energy, takes place when a positron combines with an electron to produce two photons, called the annihilation radiation.

Interactions of charged particles: Charged particles (electrons, protons, α particles, and nuclei) interact principally by ionization and excitation. Radiative collisions in which the charged particle interacts by the bremsstrahlung process are possible but are much more likely for electrons than for heavier charged particles.

The charged particle interactions or collisions are mediated by coulomb force between the electric field of the traveling particle and electric fields of orbital electrons and nuclei of atoms of the material. Collisions between the particle and the atomic electrons result in ionization and excitation of the atoms. Collisions between the particle and the nucleus result in radiative loss of energy or bremsstrahlung. Particles also suffer scattering without significant loss of energy. Because of much smaller mass, electrons suffer greater multiple scattering than do heavier particles.

The rate of kinetic energy loss per unit path length of the particle (dE/dx) is known as the stopping power (S). The quantity S/r is called the mass stopping power, where r is the density of the medium and is usually expressed in MeV cm^2/g.

Interactions of electrons: Interactions of electrons when passing through matter are quite similar to those of heavy particles. However, because of their relatively small mass, the electrons suffer greater multiple scattering and changes in direction of motion. As a consequence, the Bragg peak is not observed for electrons. Multiple changes in direction during the slowing down process smears out the Bragg peak.

In water or soft tissue, electrons, like other charged particles, lose energy predominantly by ionization and excitation. This results in deposition of energy or absorbed dose in the medium. In the process of ionization, occasionally the stripped electron receives sufficient energy to produce an ionization track of its own. This ejected electron is called a secondary electron, or a δ ray.

Because of its small mass, an electron may interact with the electromagnetic field of a nucleus and be decelerated so rapidly that a part of its energy is lost as bremsstrahlung. The rate of energy loss as a result of bremsstrahlung increases with the increase in the energy of the electron and the atomic number of the medium.

Interactions of neutrons:

Neutrons interact basically by two processes: (a) recoiling protons from hydrogen and recoiling heavy nuclei from other elements, and (b) nuclear disintegrations.

The first process may be likened to a billiard-ball collision in which the energy is redistributed after the collision between the colliding particles. The energy transfer is very efficient if the colliding particles have the same mass (e.g., a neutron colliding with a hydrogen nucleus). On the other hand, the neutron loses very little energy when colliding with a heavier nucleus. Thus, the most efficient absorbers of a neutron beam are the hydrogenous materials such as paraffin wax or polyethylene. Lead, which is a very good absorber for x-rays, is a poor shielding material against neutrons.

Dose deposited in tissue from a high-energy neutron beam is predominantly contributed by recoil protons. Because of the higher hydrogen content, the dose absorbed in fat exposed to a neutron beam is about 20% higher than in muscle. Nuclear disintegrations produced by neutrons result in the emission of heavy charged particles, neutrons, and γ rays and give rise to about 30% of the tissue dose.

MEASUREMENT OF IONISING RADIATION

Roentgen: The ICRU (1) defines exposure (X) as the quotient of dQ by dm where dQ is the absolute value of the total charge of the ions of one sign produced in air when all the electrons (negatrons and positrons) liberated by photons in air of mass dm are completely stopped in air.

$$1\ \mathrm{R} = 2.58 \times 10^{-4}\ \mathrm{C/kg\ air}$$

Exposure applies only to x and γ radiations, is a measure of ionization in air only, and cannot be used for photon energies above about 3 MeV.

According to the definition of roentgen, the electrons produced by photons in a specified volume must spend all their energies by ionization in air enclosed by the plates (region of ion collection) and the total ionic charge of either sign should be measured. However, some electrons produced in the specified volume deposit their energy outside the region of ion collection and thus are not measured. On the other hand, electrons produced outside the specified volume may enter the ion-collecting region and produce ionization there. If the ionization loss is compensated by the ionization gained, a condition of **electronic equilibrium** exists. Under this condition, the definition of roentgen is effectively satisfied. This is the principle of the free-air ionization chamber, a primary standard used only for the calibration of secondary instruments designed for field use.

There are limitations on the design of a free-air chamber for the measurement of roentgens for high-energy x-ray beams. As the photon energy increases, the range of the electrons liberated in air increases rapidly. This necessitates an increase in the separation of the plates to maintain electronic equilibrium. Too large a separation, however, creates problems of nonuniform electric field and greater ion recombination. Although the plate separation can be reduced by using air at high pressures, the problems still remain in regard to air attenuation, photon scatter, and reduction in the efficiency of ion collection. Because of these problems, there is an upper limit on the photon energy above which the roentgen cannot be accurately measured. This limit occurs at about **3 MeV**.

Thimble chamber:

In the 100- to 250-kVp x-ray range, the wall thickness of the thimble (assuming unit density) is about 1 mm, and in the case of ^{60}Co γ rays (average hv ≈ 1.25 MeV), it is approximately 5 mm. In practice, however, a thimble chamber is constructed with wall thicknesses of 1 mm or less and this is supplemented with close-fitting caps of Plexiglas or other plastic to bring the total wall thickness up to that needed for electronic equilibrium for the radiation in question.

The wall is shaped like a sewing thimble. The inner surface of the thimble wall is coated by a special material to make it electrically conducting. This forms one electrode. The other electrode is a rod of low-atomic-number material such as graphite or aluminum held in the center of the thimble but electrically insulated from it. A suitable voltage is applied between the two electrodes to collect the ions produced in the air cavity.

For the thimble chamber to be air equivalent, the effective atomic number of the wall material and the central electrode must be such that the system as a whole behaves like a free-air chamber. Most commonly used wall materials are made either of graphite (carbon), Bakelite, or a plastic coated on the inside by a conducting layer of graphite or of a conducting mixture of Bakelite and graphite.

Measurement of exposure:

Exposure in units of roentgen can be measured with a thimble chamber having an exposure calibration factor N_C traceable to the National Institute of Standards and Technology (NIST) for a given quality of radiation. The chamber is held at the desired point of measurement in the same configuration as used in the chamber calibration (Fig. 6.7). Precautions are taken to avoid media, other than air, in the vicinity of the chamber that might scatter radiation. Suppose a reading M is obtained for a given exposure. This can be converted to roentgens as follows:

$$X = M \cdot N_C \cdot C_{T,P} \cdot C_s \cdot C_{st} \qquad (6.13)$$

where $C_{T,P}$ is the correction for temperature and pressure (Equation 6.12), C_s is the correction for loss of ionization as a result of recombination (section 6.8), and C_{st} is the stem leakage correction (section 6.5). The quantity X given by Equation 6.13 is the exposure that would be expected in free air at the point of measurement in the absence of the chamber. In other words, the correction for any perturbation produced in the beam by the chamber is inherent in the chamber calibration factor N_C.

- Chambers that require calibration by a standard chamber are called secondary chambers (e.g., condenser chambers, Farmer chambers).
- Condenser chambers (e.g., Victoreen R meters) may be used for exposure measurements up to cobalt-60. They are not suitable for higher-energy beams.
- Farmer or Farmer-type chambers can be used to calibrate all beam energies used in therapy.
- Extrapolation and parallel-plate (or plane-parallel) chambers are suitable for measuring surface dose or dose in the buildup region where dose gradients are high.

QUALITY OF X RAY BEAMS

In the case of **low**-energy x-ray beams (below megavoltage range), it is customary to describe quality in terms of HVL together with **kVp**, although HVL alone is adequate for most clinical applications.

On the other hand, in the **megavoltage** x-ray range, the quality is specified by the **peak energy** and rarely by the HVL. The reason for this convention is that in the megavoltage range the beam is so heavily filtered through the transmission-type target and the flattening filter that any additional filtration does not significantly alter the beam quality or its HVL. Thus, for a "hard" beam with a fixed filtration, the x-ray energy spectrum is a function primarily of the peak energy and so is the beam quality.

The a**verage energy** of such a beam is approximately **one third of the peak energy**.

Thoraeus filter:

Combination filters containing plates of tin, copper, and aluminum have been designed to increase the resulting half-value layer of the orthovoltage beams without reducing the beam intensity to unacceptably low values. Such filters are called Thoraeus filters . It is important that the combination filters be arranged in the proper order, with the highest-atomic-number material nearest the x-ray target. Thus, a Thoraeus filter is inserted with **tin facing the x-ray tube** and the **aluminum facing the patient**, with the copper sandwiched between the tin and the aluminum plates.

Filter	Composition
Thoraeus I	0.2 mm Sn + 0.25 mm Cu + 1 mm Al
Thoraeus II	0.4 mm Sn + 0.25 mm Cu + 1 mm Al
Thoraeus III	0.6 mm Sn + 0.25 mm Cu + 1 mm Al

For cesium and cobalt teletherapy machines, filters are not needed because the beams are almost monoenergetic. Although a megavoltage x-ray beam has a spectrum of energies, the beam is hardened by the inherent filtration of the transmission target as well as by transmission through the flattening filter. Thus, no additional filtration is required to improve the beam quality.

- Peak voltage (kVp) applied to an x-ray generator can be measured directly (e.g., voltage divider, sphere-gap method) or indirectly (e.g., fluorescence, attenuation, or a penetrameter device such as an Adrian-Crooks cassette).
- Peak energy (MV) of a megavoltage x-ray beam can be measured directly by scintillation spectrometry or by photoactivation of appropriate foils (e.g., PAR method). Most commonly used methods, however, are indirect, such as comparing measured percent depth dose distribution in water with published data.
- Energy spectrum of an x-ray beam can be measured by scintillation spectrometry. The spectrum may be displayed in terms of photon fluence per unit energy interval as a function of photon energy.

MEASUREMENT OF ABSORBED DOSE

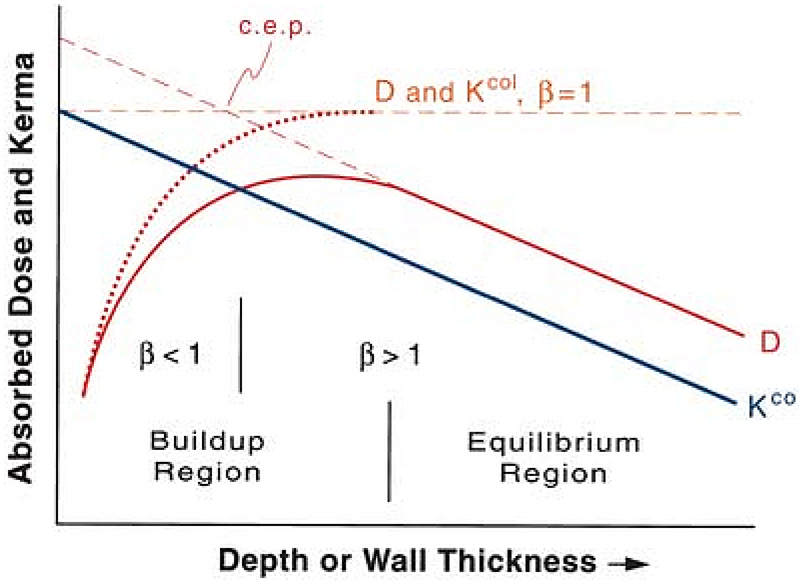
The quantity kerma (K) (kinetic energy released in the µedium) is defined as “the quotient of dE_{tr} by dm, where dE_{tr} is the sum of the initial kinetic energies of all the charged ionizing particles (electrons and positrons) liberated by uncharged particles (photons) in a material of mass dm”

For a photon beam traversing a medium, kerma at a point is directly proportional to the photon energy fluence ψ and is given by:

$$K = \Psi\left(\frac{\bar{\mu}_{tr}}{\rho}\right)$$

Exposure is the ionization equivalent of the collision kerma in air

Whereas kerma is maximum at the surface and decreases with depth, the dose initially builds up to a maximum value and then decreases at the same rate as kerma.



$$D_{\text{air}} = (K^{\text{col}})_{\text{air}} = X \cdot \frac{\overline{W}}{e}$$

Roentgen-to-rad conversion factor=0.876 rad/R

Bragg-Gray Cavity Theory:

Calculation of absorbed dose from exposure is subject to some major limitations. For instance, it may not be used for photons above 3 MeV and may not be used in cases where electronic equilibrium does not exist. In addition, the term exposure applies only to x and γ radiations and not valid for particle dosimetry. The Bragg-Gray cavity theory, on the other hand, may be used without such restrictions to calculate dose directly from ion chamber measurements in a medium.

According to the Bragg-Gray theory, the ionization produced in a gas-filled cavity placed in a medium is related to the energy absorbed in the surrounding medium. When the cavity is sufficiently small so that its introduction into the medium does not alter the number or distribution of the electrons that would exist in the medium without the cavity, then the following Bragg-Gray relationship is satisfied:

$$D_{\text{med}} = J_g \cdot \frac{\overline{W}}{e} \cdot (\overline{S}/\rho)_g^{\text{med}}$$

- All recent calibration protocols (TG-21, TG-51, and IAEA TRS-398) use B-G cavity theory.
- The major upgrade of TG-51 over TG-21 is the chamber calibration, which is based on absorbed dose–to–water instead of exposure in air.
- Beam quality for photon beams is specified by percent depth dose for the photon component of the beam at 10 cm depth in water ($\%dd(10)_x$).
- Beam quality for the purpose of electron beam calibration is specified by the depth of 50% dose in water (R_{50}).

- **Exposure rate constant** is defined as exposure rate from a radioactive source of point size and unit activity at a unit distance. Its unit is $Rm^2h^{-1}Ci^{-1}$, which means roentgens per hour at a distance of 1 meter from a point source of activity of 1 Ci.
- Absolute dosimeters
  - Absolute dosimetry means that the dose is determined from the first principles —without reference to another dosimeter.
  - The free-air ionization chamber, specially designed spherical chambers of known volume (e.g., at NIST), the calorimeter, and the ferrous sulfate (Fricke) dosimeter are examples of absolute dosimeters. They are also called primary standards.
- Secondary dosimeters
  - Secondary dosimeters require calibration against a primary standard. Examples are thimble chambers and plane-parallel ion chambers. Thermoluminescent dosimeters, diodes, and film are also secondary dosimeters but are used primarily for relative dosimetry. They require calibration against a calibrated ion chamber as well as appropriate corrections for energy dependence (e.g., with depth) and other conditions that may affect their dose response characteristics.
- Radiographic films: Film is well suited for relative dosimetry of electron beams (shows practically no energy dependence). In photon beams, however, it shows significant

energy dependence and therefore it is used mostly for portal imaging and quality assurance procedures such as checking beam alignment, isocentric accuracy, and beam flatness. For measuring dose distributions, photon energy dependence must be taken into account.

- Radiochromic films: Major advantages include almost tissue equivalence, high spatial resolution, large dynamic range (10^{-2}–10^{6} cGy), low energy dependence, insensitivity to visible light, and no need for processing. It is well suited for dosimetry of brachytherapy sources where the doses and dose gradients close to the sources are very high.

DOSE DISTRIBUTION & SCATTER ANALYSIS

Phantoms:

Since it is not always possible to put radiation detectors in water, solid dry phantoms have been developed as substitutes for water. Ideally, for a given material to be tissue or water equivalent, it must have the **same effective atomic number, number of electrons per gram,** and **mass density**. However, since the Compton effect is the most predominant mode of interaction for megavoltage photon beams in the clinical range, the necessary condition for water equivalence for such beams is the **same electron density** (number of electrons per cubic centimeter) as that of water. (=3.34 x 10^{23} /gm)
Common materials include plexiglass, polystyrene, paraffin, polyethylene & solid water.

PDD:

Percentage (or simply percent) depth dose may be defined as the quotient, expressed as a percentage, of the absorbed dose at any depth d to the absorbed dose at a fixed reference depth d_0, along the central axis of the beam. For higher energies, the reference depth is taken at the position of the peak absorbed dose ($d_0 = d_m$)
PDD depends on beam quality, field size and shape, depth, SSD.

Physics of build-up region:

(1.a) As the high-energy photon beam enters the patient or the phantom, high-speed electrons are ejected from the surface and the subsequent layers. (b) These electrons deposit their energy a significant distance away from their site of origin. (c) Because of (a) and (b), the electron fluence and hence the absorbed dose increase with depth until they reach a maximum. However, the photon energy fluence continuously decreases with depth and, as a result, the production of electrons also decreases with depth. The net effect is that beyond a certain depth the dose eventually begins to decrease with depth.

Geometric field size is defined as “the projection, on a plane perpendicular to the beam axis, of the distal end of the collimator as seen from the front center of the source”.

The dosimetric, or the physical, field size is the distance intercepted by a given isodose curve (usually 50% isodose) on a plane perpendicular to the beam axis at a stated distance from the source.

Effect of field size on PDD:

For a sufficiently small field one may assume that the depth dose at a point is effectively the result of the primary radiation, that is, the photons that have traversed the overlying medium without interacting. The contribution of the scattered photons to the depth dose in this case is negligibly small or 0. But as the field size is increased, the contribution of the scattered radiation to the absorbed dose increases. Because this **increase in scattered dose** is greater at larger depths than at the depth of D_{max}, the percent depth dose **increases** with increasing field size.

The field size dependence of percent depth dose is less pronounced for the higher-energy than for the lower-energy beams.

Effect of SSD on PDD:

Although the actual dose rate at a point decreases with an increase in distance from the source, the percent depth dose, which is a relative dose with respect to a reference point, increases with SSD. the drop in dose rate between two points is much greater at smaller distances from the source than at large distances. This means that the percent depth dose, which represents depth dose relative to a reference point, decreases more rapidly near the source than far away from the source.

The Mayneord F factor method is based on a strict application of the inverse square law, without considering changes in scattering, as the SSD is changed. The Mayneord F factor method works reasonably well for small fields since the scattering is minimal under these conditions. However, the method can give rise to significant errors under extreme conditions such as lower energy, large field, large depth, and large SSD change.

In general, the Mayneord F factor overestimates the increase in percent depth dose with increase in SSD.

TAR:

Tissue-air ratio may be defined as the ratio of the dose (D_d) at a given point in the phantom to the dose in free space (D_{fs}) at the same point. It does not depend on SSD.

BSF:

The term backscatter factor (BSF) is simply the tissue-air ratio at the depth of maximum dose on central axis of the beam. It may be defined as the ratio of the dose on central axis at the depth of maximum dose to the dose at the same point in free space. The backscatter factor, like the tissue-air ratio, is independent of distance from the source and depends only on the beam quality and field size. Whereas BSF increases with field size, its maximum value occurs for beams having a half-value layer between 0.6 and 0.8 mm Cu, depending on field size. For megavoltage beams (^{60}Co and higher energies), the backscatter factor is much smaller. This increase in dose is the result of radiation scatter reaching the point of D_{max} from the overlying and underlying tissues. As the beam energy is increased, the scatter is further reduced and so is the backscatter factor.

Relation between BSF, PDD, TAR:

$$P(d,r,f) = \mathrm{TAR}(d,r_d) \cdot \frac{1}{\mathrm{BSF}(r)} \cdot \left(\frac{f+d_m}{f+d}\right)^2 \cdot 100$$

SAR:

Scatter-air ratios are used for the purpose of calculating scattered dose in the medium. The computation of the primary and the scattered dose separately is particularly useful in the dosimetry of irregular fields.

Scatter-air ratio may be defined as the ratio of the scattered dose at a given point in the phantom to the dose in free space at the same point. The scatter-air ratio, like the tissue-air ratio, is independent of the source to surface distance but depends on the beam energy, depth, and field size. Because the scattered dose at a point in the phantom is equal to the total dose minus the primary dose at that point, scatter-air ratio is mathematically given by the difference between the TAR for the given field and the TAR for the 0 × 0 field:

$$\mathrm{SAR}(d,r_d) = \mathrm{TAR}(d,r_d) - \mathrm{TAR}(d,0)$$

SAR is used for dosimetry of irregularly shaped fields, by Clarkson's method.

A SYSTEM OF DOSIMETRIC CALCULATION

The dose to a point in a medium may be analyzed into primary and scattered components. The **primary** dose is contributed by the initial or original photons emitted from the source and the **scattered** dose is the result of the scattered photons. The scattered dose can be further analyzed into collimator and phantom components, because the two can be varied independently by blocking.

For megavoltage photon beams, it is reasonably accurate to **consider collimator scatter as part of the primary beam** so that the phantom scatter could be calculated separately. Therefore, we define an effective primary dose as the dose due to the primary photons as well as those scattered from the collimating system. The effective primary in a phantom may be thought of as the dose at depth minus the phantom scatter. Alternatively, the effective primary dose may be defined as the depth dose expected in the field when scattering volume is reduced to zero while keeping the collimator opening, hence the concept of a **0 × 0 field to represent primary beams** with the implicit assumption that lateral electronic equilibrium exists at all points .

Collimator scatter factor:

The beam output (exposure rate, dose rate in free space, or energy fluence rate) measured in air depends on the field size. As the field size is increased, the output increases because of the **increased collimator scatter**,w hich is added to the primary beam.

For photon beams for which backscatter factors can be accurately measured (e.g., up to cobalt-60), S_p factor at the depth of maximum dose may be defined simply as the ratio of backscatter factor (BSF) for the given field to that for the reference field.

TPR:

The TPR is defined as the ratio of the dose at a given point in phantom to the dose at the same point at a fixed reference depth, usually **5 cm**.

TMR:

TMR is a special case of TPR and may be defined as the ratio of the dose at a given point in phantom to the dose at the same point at the reference depth of maximum dose.

SMR:

SMR is the ratio of the scattered dose at a given point in phantom to the effective primary dose at the same point at the reference depth of maximum dose. For ^{60}Co γ rays, SMRs are approximately the same as SARs.

A. Accelerator Calculations

A.1. SSD Technique

Percent depth dose is a suitable quantity for calculations involving SSD techniques. Machines are usually calibrated to deliver 1 rad (10^{-2} Gy) per monitor unit (MU) at the reference depth t_0, for a reference field size 10 × 10 cm and a source to calibration point distance of SCD. Assuming that the S_c factors relate to collimator field sizes defined at the SAD, the monitor units necessary to deliver a certain tumor dose (TD) at depth d for a field size r at the surface at any SSD are given by:

$$\mathrm{MU} = \frac{\mathrm{TD} \times 100}{\mathrm{K} \times (\%\mathrm{DD})_d \times S_c(r_c) \times S_p(r) \times (\mathrm{SSD\ factor})} \qquad (10.8)$$

where K is 1 rad per MU, r_c is the collimator field size, given by:

$$r_c = r \cdot \frac{\mathrm{SAD}}{\mathrm{SSD}}$$

$$\mathrm{SSD\ factor} = \left(\frac{\mathrm{SCD}}{\mathrm{SSD} + t_0}\right)^2$$

A.2. Isocentric Technique

TMR is the quantity of choice for dosimetric calculations involving isocentric techniques. Since the unit is calibrated to give 1 rad (10^{-2} Gy)/MU at the reference depth t_0 and calibration distance SCD and for the reference field (10 × 10 cm), then the monitor units necessary to deliver isocenter dose (ID) at depth d are given by:

$$\mathrm{MU} = \frac{\mathrm{ID}}{\mathrm{K} \times \mathrm{TMR}(d, r_d) \times S_c(r_c) \times S_p(r_d) \times \mathrm{SAD\ factor}} \qquad (10.9)$$

P.164

where:

$$\mathrm{SAD\ factor} = \left(\frac{\mathrm{SCD}}{\mathrm{SAD}}\right)^2$$

Off-axis correction:

The primary dose distribution has been shown to vary with lateral distance from the central axis because of the change in beam quality, as mentioned earlier. Therefore, the percent depth dose or TMR distribution along the central ray of an asymmetric field is not the same as along the central axis of a symmetric field of the same size and shape. In addition, the incident primary beam fluence at off-axis points varies as a function of distance from the central axis, depending on the flattening filter design. These effects are not emphasized in the dosimetry of symmetric fields, because target doses are usually specified at the beam central axis and the off-axis dose distributions are viewed from the isodose curves. In asymmetric fields, however, the target or the point of interest does not lie on the beam central axis; therefore, an off-axis dose correction may be required in the calculation of target dose. This correction will depend on the depth and the distance from the central axis of the point of interest.

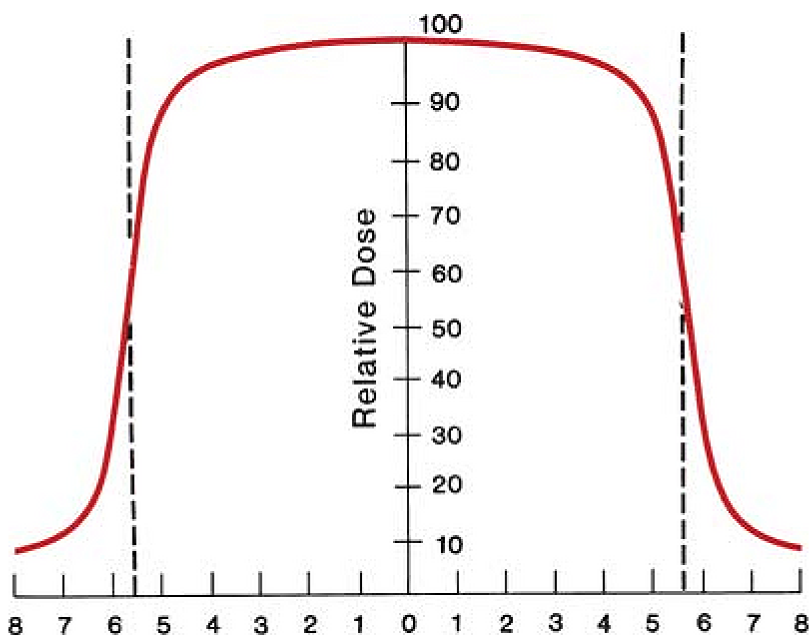
It is possible to calculate depth dose distributions at any point within the field or outside the field using Clarkson's technique

Point under the block:

The dose distribution in a blocked field is best determined by Clarkson's method of irregular field dosimetry. However, if the blocked portion of the field is approximated to a rectangle, a simpler method known as **negative field method** may be used. In this method, the dose at any point is equal to the dose from the overall (unblocked) field minus the dose expected if the entire field were blocked, leaving the shielded volume open. In other words, the blocked portion of the field is considered a negative field and its contribution is subtracted from the overall field dose distribution.

A computerized negative field method not only is a fast method of calculating isodose distribution in blocked fields, but is also very convenient for manual point dose calculation.

- The dose at any depth is greatest on the central axis of the beam and gradually decreases toward the edges of the beam, with the exception of some linac x-ray beams, which exhibit areas of high dose or “horns” near the surface in the periphery of the field. These horns are created by the flattening filter, which is usually designed to overcompensate near the surface in order to obtain flat isodose curves at greater depths.
- Outside the geometric limits of the beam and the penumbra, the dose variation is the result of side scatter from the field and both leakage and scatter from the collimator system. Beyond this collimator zone, the dose distribution is governed by the lateral scatter from the medium and leakage from the head of the machine (often called therapeutic housing or source housing).



The dose variation across the field at a specified depth. Such a representation of the beam is known as the beam profile.

The ionization chamber used for isodose measurements should be small so that measurements can be made in regions of high dose gradient, such as near the edges of the beam. It is recommended that the sensitive volume of the chamber be less than 15 mm long and have an inside diameter of 5 mm or less. Energy independence of the chamber is another important requirement. Because the x-ray beam spectrum changes with position in the phantom owing to scatter, the energy response of the chamber should be as flat as possible. This can be checked by obtaining the exposure calibration of the chamber for orthovoltage (1–4 mm Cu) and ^{60}Co beams. A variation of approximately **5%** in response throughout this energy range is acceptable.

The wedge tray is always at a distance of at least 15 cm from the skin surface, so as to avoid destroying the skin-sparing effect of the megavoltage beam.

Single reference depth of 10 cm for wedge angle specification.

In cobalt-60 teletherapy, the wedge factor is sometimes incorporated into the isodose curves. If such a chart is used for isodose planning, no further correction should be applied to the output. In other words, the machine output corresponding to the open beam should be used.

The wedge filter alters the beam quality by preferentially attenuating the lower-energy photons (beam hardening) and, to a lesser extent, by Compton scattering, which results in energy degradation (beam softening). For the ^{60}Co beam, because the primary beam is essentially monoenergetic, the presence of the wedge filter does not significantly alter the central axis percent depth dose distribution. For x-rays, on the other hand, there can be some beam hardening, and consequently, the depth dose distribution can be somewhat altered, especially at large depths.

As the patient thickness increases or the beam energy decreases, the central axis maximum dose near the surface increases relative to the midpoint dose. This effect is called **tissue lateral effect.**

For parallel opposed beams, they have shown that treating with one field per day produces greater biologic damage to normal subcutaneous tissue than treating with two fields per day, despite the fact that the total dose is the same. Apparently, the biologic effect in the normal tissue is greater if it receives alternating high- and low-dose fractions compared with the equal but medium-size dose fractions resulting from treating both fields daily. This phenomenon has been called the edge effect, or the tissue lateral damage.

Integral dose is a measure of the total energy absorbed in the treated volume. If a mass of tissue receives a uniform dose, then the integral dose is simply the product of mass and dose.

$$\theta = 90^\circ - \Phi/2$$

The relationship assumes that the wedge isodose curves are not modified by the **surface contour**. In practice, however, contours are usually curved or irregular in shape and thus modify the isodose distribution for the wedged beams. As a result, the isodose curves for the individual fields are no longer parallel to the bisector of the hinge angle, thus giving rise to a nonuniform distribution in the overlap region. This problem can be solved by using compensators, which make the skin surface effectively flat and perpendicular to each beam. An alternative approach is to modify the wedge angle (using a different wedge angle filter) so that a part of the wedge angle acts as a compensator and the rest as a true wedge filter. The main objective is to make the isodose curves parallel to the hinge angle bisector.

The volume enclosed by an isodose surface that represents the minimum target dose that adequately covers the PTV is called the **treated volume**.

The volume of tissue receiving a significant dose (e.g., ≥50% of the specified target dose) is called the **irradiated volume**.

TREATMENT PLANNING: PATIENT DATA, CORRECTIONS & SETUP

CT

The reconstruction algorithm generates what is known as CT numbers, which are related to attenuation coefficients. The CT numbers range from –1,000 for air to +1,000 for bone, with that for water set at 0. The CT numbers normalized in this manner are called Hounsfield numbers (H):

$$H = \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}} \times 1{,}000$$

where μ is the linear attenuation coefficient. Thus, a Hounsfield unit represents a change of 0.1% in the attenuation coefficient of water.

Because the CT numbers bear a linear relationship with the attenuation coefficients, it is possible to infer electron density (electrons cm^{-3})

MRI

Basic physics of MRI involves a phenomenon known as nuclear magnetic resonance (NMR). It is a resonance transition between nuclear spin states of certain atomic nuclei when subjected to a radiofrequency (RF) signal of a specific frequency in the presence of an external magnetic field.

The nuclei that participate in this phenomenon are the ones that intrinsically possess spinning motion (i.e., have angular momentum). These rotating charges act as tiny magnets with associated magnetic dipole moment, a property that gives a measure of how quickly the magnet will align itself along an external magnetic field. Because of the spinning motion or the magnetic dipole moment, nuclei align their spin axes along the external magnetic field (H) as well as orbit or precess around it. The frequency of precession is called the Larmor frequency. A second alternating field is generated by applying an alternating voltage (at the Larmor frequency) to an RF coil. This field is applied **perpendicular t**o H and rotates around H at the Larmor frequency. This causes the nuclei to precess around the new field in the transverse direction. When the RF signal is turned off, the nuclei return to their original alignment around H. This transition is called **relaxation**. It induces a signal in the receiving RF coil (tuned to the Larmor frequency), which constitutes the NMR signal.

The turning off of the transverse RF field causes nuclei to relax in the transverse direction (T_2 relaxation) as well as to return to the original longitudinal direction of the magnetic field (T_1 relaxation). The relaxation times, T_1 and T_2, are actually time constants (like the decay constant in radioactive decay) for the exponential function that governs the two transitions.

Most MR imaging uses a spin echo technique in which a 180-degree RF pulse is applied after the initial 90-degree pulse, and the resulting signal is received at a time that is equal to twice the interval between the two pulses. This time is called the echo time (TE). The time between each 90-degree pulse in an imaging sequence is called the repetition time (TR). By adjusting TR and TE, image contrast can be affected.

For example, a long TR and short TE produces a proton (spin) density-weighted image, a short TR and a short TE produces a T_1-weighted image, and a long TR and a long TE produces a T_2-weighted image.

USG

An ultrasonic transducer converts electrical energy into ultrasound energy, and vice versa. This is accomplished by a process known as the piezoelectric effect. Most common crystals used clinically are made artificially such as barium titanate, lead zirconium titanate, and lead metaniobate.

Dosimetric corrections for surface irregularities:

(2) Isodose shift method
(3) TAR method
(4) Effective SSD method

Dose at interfaces: Beyond the interface, dose is increased after air, decreased after bone.Before the interface, due to differential backscatter, dose is reduced in case of air and increased in case of bone.

Dose to soft tissue within bone is two to five times higher in the orthovoltage and superficial range of beam energies. It is about 3% to 10% higher in the megavoltage range used clinically

Compensator design:

Because the compensator is designed to be positioned at a distance from the surface, the dimensions and shape of the compensator must be adjusted because of (a) the beam divergence, (b) the relative linear attenuation coefficients of the filter material and soft tissues, and (c) the reduction in scatter at various depths when the compensator is placed at a distance from the skin rather than in contact with it. To compensate for this scatter, the compensator is designed such that the attenuation of the filter is **less** than that required for primary radiation only.

The required thickness of a tissue-equivalent compensator along a ray divided by the missing tissue thickness along the same ray may be called the density ratio or thickness ratio.

The concept of thickness ratios also reveals that a compensator cannot be designed to provide absorbed dose compensation exactly at all depths. If, for given irradiation conditions, τ is chosen for a certain compensation depth, the compensator overcompensates at shallower depths and undercompensates at greater depths. Considering the limitations of the theory and too many variables affecting τ, we have found that an average value of **0.7** for τ may be used for all irradiation conditions provided d greater than or equal to 20 cm.

The concepts of compensator ratio and the thickness ratio are the same, except that the two quantities are inverse of each other.

One simple way of constructing a two-dimensional compensator is to use thin sheets of lead (with known thickness ratio or effective attenuation coefficient) and gluing them together in a stepwise fashion to form a laminated filter. The total thickness of the filter at any point is calculated to compensate for the air gap at the point below it.

Compensating wedge vs wedge filter:

Distinction needs to be made between a wedge filter and a compensating wedge. Although a wedge filter can be used effectively as a compensator, it is primarily designed to tilt the standard isodose curves The wedge filter isodose curves must be available and used to obtain the composite isodose curves before the filter is used in a treatment setup. The C-wedge, on the other hand, is used just as a compensator, so that the standard isodose charts can be used without modification. In addition, no wedge transmission factors are required for the C-wedges.

An important advantage of C-wedges over wedge filters used as compensators is that the C-wedges can be used for partial field compensation; that is, the C-wedge is used to compensate only a part of the contour, which is irregular in shape. A wedge filter, in this case, could not be used as a compensator because it is designed to be placed in the field in a fixed position.

TREATMENT PLANNING:FIELD SHAPING, SKIN DOSE & FIELD SEPERATION

- 5 HVLs allow 3.125% transmission. Thickness of lead required to give 5% primary beam transmission is 4.3 half-value layer.
- Half-beam blocking gives rise to tilting of the isodose curves toward the blocked edge. This effect is due to missing electron and photon scatter from the blocked part of the field into the open part of the field.
- Surface dose in megavoltage beams is predominantly due to the electron contamination of the incident photon beam.
- Dose at the surface or in the buildup region is best measured with an extrapolation or a plane-parallel chamber.
- Surface dose depends on beam energy, field size, SSD, and tray to surface distance.
- Electron filters are medium-atomic-number absorbers (Z ~50) that reduce the surface dose by scattering contaminant electrons more than generating them.

Block thickness for<5% transmission:

^{137}Cs	3.0 cm
^{60}Co	5.0 cm
4 MV	6.0 cm
6 MV	6.5 cm
10 MV	7.0 cm

Cerrobend:

Lipowitz metal (brand name, Cerrobend), which has a density of 9.4 g/cm^3 at 20°C (~83% of lead density). This material consists of 50.0% bismuth, 26.7% lead, 13.3% tin, and 10.0% cadmium . In the megavoltage range of photon beams, the most commonly used thickness is 7.5 cm, which is equivalent to about 6 cm of pure lead.

MLCs:

Typical MLC systems consist of 80 leaves (40 pairs) or more. The individual leaf has a width of 1 cm or less as projected at the isocenter. The leaves are made of tungsten alloy (ρ = 17.0–18.5 g/cm^3) and have thickness along the beam direction ranging from 6 cm to 7.5 cm, depending on the type of accelerator. The leaf thickness is sufficient to provide primary x-ray transmission through the leaves of less than **2%** (compared with about 1% for jaws and 3.5% for Cerrobend blocks). The interleaf (between sides) transmission is usually less than **3%**.

The best orientation of the collimator is when the direction of motion of the leaves is parallel with the direction in which the target volume has the smallest cross section

Dmax depth:

Cobalt 5mm, 4MV 1cm, 6MV 1.5 cm, 10MV 2.5 cm, 25MV 3cm.

Effect of blocks on skin dose:

When an absorber of thickness greater than the range of secondary electrons (equilibrium thickness) is introduced in the beam, the collimator electrons are almost completely absorbed but the absorber itself becomes the principal source of electron contamination of the beam. By increasing the distance between the tray and the surface, the electron fluence incident on the skin is reduced because of divergence as well as absorption and scattering of electrons in the air. For small fields an air gap of **15**

to 20 cm between the scatterer and the skin is adequate to keep the skin dose to an acceptable level (<50% of the D_{max}).

Effect of field size on skin dose:

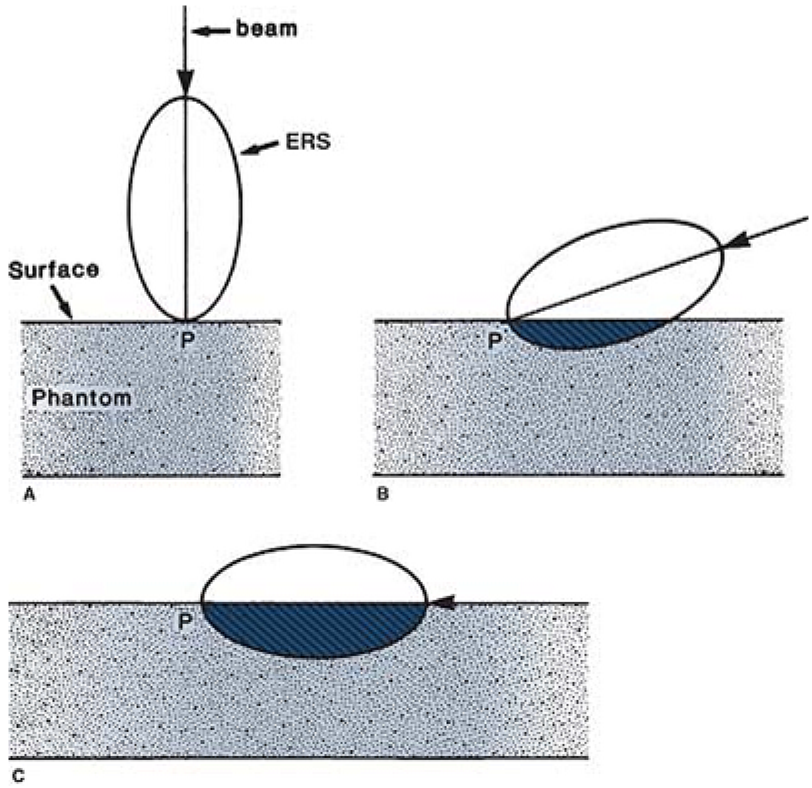
The relative skin dose depends strongly on **field size**. As the field dimensions are increased, the dose in the buildup region increases. This increase in dose is due to increased electron emission from the collimator and air.

Effect of oblique incidence on skin dose:

The ERS is a 3-D representation of secondary electron range and distribution produced by a pencil beam of photons interacting with the medium. Electrons generated inside the ERS volume will reach P and contribute to the dose there, whereas those generated outside, because of their inadequate range, make no contribution. The increase in the angle of incidence of the photon beam results in additional surface dose at P because of electron contribution from the portion of the ERS, which appears below the phantom surface. For tangential beam incidence, since half of the ERS is below the phantom surface, an upper estimate of the dose to the skin may be obtained by the following relationship:

$$\text{Percent skin dose} = \frac{1}{2}(100\% + \text{entrance dose})$$

Another important effect associated with oblique angles is that as the surface dose increases with the angle of incidence, the depth of maximum buildup decreases. The dose reaches its maximum value faster at glancing angles than at normal incidence. As a result, the dose buildup region is compressed into a more superficial region. Under these conditions, a high skin reaction becomes much more likely.



Surface separation of 2 overlapping photon fields:

$$S = S_1 + S_2 = \frac{1}{2} \cdot L_1 \cdot \frac{d}{\mathrm{SSD}_1} + \frac{1}{2} \cdot L_2 \cdot \frac{d}{\mathrm{SSD}_2}$$

Surface separation of 2 orthagonal photon fields:

$$S = \frac{1}{2} \cdot L \cdot \frac{d}{\mathrm{SSD}}$$

Orthagonal field matching for CSI:
Collimator rotation of cranial fields to eliminate cranial divergence of spinal field.
Couch rotation to eliminate caudal divergence of cranial fields, towards the direction of the cranial field.

$$\theta_{\mathrm{coll}} = \mathrm{arc\ tan}\left(\frac{1}{2} \cdot L_1 \cdot \frac{1}{\mathrm{SSD}}\right)$$

$$\theta_{\mathrm{couch}} = \mathrm{arc\ tan}\left(\frac{1}{2} \cdot L_1 \cdot \frac{1}{\mathrm{SSD}}\right)$$

RADIATION PROTECTION

- The dosimetric quantity relevant to radiation protection is the dose equivalent (=dose × quality factor).

Radiation	*Quality Factor*
X-rays, γ rays, and electrons	1
Thermal neutrons	5
Neutrons, heavy particles	20

- Effective dose equivalent is "the sum of the weighted dose equivalents for individual tissues or organs."

Tissue (T)	*Risk Coefficient*	W_T
Gonads	40×10^{-4} Sv^{-1} (40×10^{-6} rem^{-1})	0.25
Breast	25×10^{-4} Sv^{-1} (25×10^{-6} rem^{-1})	0.15
Red bone marrow	20×10^{-4} Sv^{-1} (20×10^{-6} rem^{-1})	0.12
Lung	20×10^{-4} Sv^{-1} (20×10^{-6} rem^{-1})	0.12
Thyroid	5×10^{-4} Sv^{-1} (5×10^{-6} rem^{-1})	0.03
Bone surface	5×10^{-4} Sv^{-1} (5×10^{-6} rem^{-1})	0.03
Remainder	50×10^{-4} Sv^{-1} (50×10^{-6} rem^{-1})	0.30
Total	165×10^{-4} Sv^{-1} (165×10^{-6} rem^{-1})	1.00

The total effective dose equivalent for a member of the population in the United States from various sources of natural background radiation is approximately 3.0 mSv/year. These include cosmic radiation, naturally occuring radionucleides –external (uranium, radon) & internal (K^{40}).

Exposures to **low-level** radiation may produce (a) genetic effects, such as radiation-induced gene mutations, chromosome breaks, and anomalies; (b) neoplastic diseases, such as increased incidence of leukemia, thyroid tumors, and skin lesions; (c) effect on growth and development, such as adverse effects on the fetus and young children; (d) effect on life span, such as diminishing of life span or premature aging; and (e) cataracts or opacification of the eye lens.

The harmful effects of radiation may be classified into two general categories: stochastic effects and nonstochastic effects. The NCRP (2) defines these effects as follows.

A **stochastic effect** is one in which "the **probability** of occurrence increases with increasing absorbed dose but the **severity** in affected individuals does not depend on the magnitude of the absorbed dose." In other words, a stochastic effect is an all-or-none phenomenon, such as the development of a cancer or genetic effect. Although the probability of such effects occurring increases with dose, their severity does not.

A **nonstochastic effect** is one "which **increases in severity** with increasing absorbed dose in affected individuals, owing to damage to increasing number of cells and tissues." Examples of nonstochastic effects are radiation-induced degenerative changes such as organ atrophy, fibrosis, lens opacification, blood changes, and decrease in sperm count.

Whereas **no threshold dose can be predicted for stochastic effects**, it is possible to set threshold limits on nonstochastic effects that are significant or seriously health impairing. However, for the purpose of radiation protection, a cautious assumption is made that "the dose-risk relationship is strictly proportional (linear) without threshold, throughout the range of dose equivalent and dose equivalent rates of importance in routine radiation protection."

NCRP recommendations on **exposure limits of radiation workers** are based on the following criteria:

(a) at low radiation levels the nonstochastic effects are essentially avoided;

(b) the predicted risk for stochastic effects should not be greater than the average risk of accidental death among workers in "safe" industries; and

(c) the ALARA principle should be followed, for which the risks are kept as low as reasonably achievable, taking into account social and economic factors.

Radiation workers are limited to an annual effective dose equivalent of 50 mSv (5 rem) and the general public is not to exceed one tenth of this value 5 mSv(0.5 rem) for infrequent exposure.

Occupational exposures:

A. Occupational exposures (annual)	
1. Effective dose-equivalent limit (stochastic effects)	50 mSv
2. Dose-equivalent limits for tissues and organs (nonstochastic effects)	
a. Lens of eye	150 mSv
b. All others (e.g., red bone marrow, breast, lung, gonads, skin, and extremities)	500 mSv
3. Guidance: cumulative exposure	10 mSv × age

Public exposures:

D. Public exposures (annual)	
1. Effective dose-equivalent limit, continuous or frequent exposure	1 mSv
2. Effective dose-equivalent limit, infrequent exposure	5 mSv
3. Remedial action recommended when:	
a. Effective dose equivalent	>5 mSv
b. Exposure to radon and its decay products	>0.007 Jhm^{-3}
4. Dose-equivalent limits for lens of eye, skin, and extremities	50 mSv
E. Education and training exposures (annual)	
1. Effective dose equivalent	1 mSv
2. Dose-equivalent limit for lens of eye, skin, and extremities	50 mSv
F. Embryo-fetus exposures	
1. Total dose-equivalent limit	5 mSv
2. Dose-equivalent limit in a month	0.5 mSv
G. Negligible individual risk level (annual) effective dose equivalent per source or practice	0.01 mSv

Pregnant women:

The pregnant woman who is a radiation worker can be considered as an occupationally exposed individual, but the fetus cannot. The total dose-equivalent limit to an embryo-fetus is **5 mSv** (0.5 rem), with the added recommendation that exposure to the fetus should not exceed **0.5 mSv** (0.05 rem) **in any 1 month**.

NIRL:

A negligible individual risk level (NIRL) is defined by the NCRP as “a level of average annual excess risk of fatal health effects attributable to irradiation, below which further effort to reduce radiation exposure to the individual is unwarranted." The NCRP also states that "the NIRL is regarded as trivial compared to the risk of fatality associated with ordinary, normal societal activities and can, therefore, be dismissed from consideration."

The concept of NIRL is applied to radiation protection because of the need for having a reasonably negligible risk level that can be considered as a threshold below which efforts to reduce the risk further would not be warranted or, in the words of the NCRP, "would be deliberately and specifically curtailed."

To avoid misinterpretation of the relationships between the NIRL, ALARA, and maximum permissible levels, the NCRP points out that the NIRL should not be thought of as an acceptable risk level, a level of significance, or a limit. Nor should it be the goal of ALARA, although it does provide a lower limit for application of the ALARA process. The ALARA principle encourages efforts to keep radiation exposure as low as reasonably achievable, considering the economic and social factors.

The annual NIRL has been set at a risk of 10^{-7}, corresponding to a dose equivalent of **0.01 mSv** (0.001 rem). This corresponds to a lifetime (70 years) risk of 0.7×10^{-5}.

Structural shield design:

<u>Controlled & noncontrolled areas</u>: For protection calculations, the dose-equivalent limit is assumed to be **0.1 rem/week** for the controlled areas and **0.01 rem/week for** the noncontrolled areas. These values approximately correspond to the annual limits of 5 rem/year and 0.5 rem/year, respectively.

A barrier sufficient to attenuate the useful beam to the required degree is called the primary barrier. The required barrier against stray radiation (leakage and scatter) is called the secondary barrier.

<u>Criteria for barrier thickness:</u>

Workload (W): Weekly dose delivered at 1 m from the source. Can be estimated by multiplying the number of patients treated per week with the dose delivered per patient at 1 m

Use Factor (U): Fraction of the operating time during which the radiation under consideration is directed toward a particular barrier.

Occupancy Factor (T): Fraction of the operating time during which the area of interest is occupied by the individual.

Distance (d): Distance in meters from the radiation source to the area to be protected. Inverse square law is assumed for both the primary and stray radiation.

For megavoltage x- and γ radiation, equivalent thickness of various materials can be calculated by comparing tenth value layers (TVLs) for the given beam energy.

For megavoltage therapy installations, the leakage barrier usually far exceeds that required for the scattered radiation, because the leakage radiation is more penetrating than the scattered radiation. For the lower-energy x-ray beams, however, the difference between the barrier thickness for the leakage and for the scattered radiation is relatively less.

A barrier designed for primary radiation provides adequate protection against leakage and scattered radiation. If a barrier is designed for stray radiation only, the thickness is computed for leakage and scattered radiations separately. If the thicknesses of the two barriers differ by at least three HVLs, the thicker of the two will be adequate. If the difference is less than three HVLs, one HVL should be added to the larger one to obtain the required secondary barrier.

Maze & door thickness:

Radiation is scattered at least twice before incidence on the door. Each Compton scatter at 90 degrees or greater will reduce the energy to 500 keV or less.

For megavoltage units, the attenuation curves for the 500-kVp x-rays may be used to determine the door shielding from multiply scattered x-rays. In most cases, the required shielding turns out to be **less than 6 mm of lead**.

Concrete barriers designed for x-ray shielding are sufficient for protection against neutrons. However, the door must be protected against neutrons that diffuse into the maze and reach the door.A longer maze (>5 m) is desirable in reducing the neutron fluence at the door. Finally, a few inches of a hydrogenous material such as polyethylene can be added to the door to thermalize the neutrons and reduce the neutron dose further.

Neutron contamination:

High-energy x-ray beams (e.g., >10 MV) are contaminated with neutrons. These are produced by high-energy photons and electrons incident on the various materials of target, flattening filter, collimators, and other shielding components.

Neutron production during electron beam therapy mode is quite small compared with that during the x-ray mode.

The neutron contamination increases rapidly as the energy of the beam is increased from 10 to 20 MV, and then remains approximately constant above this. Measurements have shown that in the 16- to 25-MV x-ray therapy mode the neutron dose equivalent along central axis is approximately **0.5%** of the x-ray dose and falls off to about 0.1% outside the field.

Geiger Muller counter:

The G-M tube is much more sensitive than the ionization chamber.

The Geiger counter can detect individual photons or individual particles that could never be observed in a ionization chamber.

However, this detector is not a dose-measuring device. Although a Geiger counter is useful for preliminary surveys to detect the presence of radiation, ionization chambers are recommended for quantitative measurement.

Because of their inherently slow recovery time (~50 to 300 μs), they can never record more than 1 count/machine pulse. Thus, a G-M counter could significantly underestimate radiation levels when used to count radiation around pulsed machines such as accelerators.

HDR BRACHYTHERAPY

Source dimension: 0.6 mm diameter & 1cm in length

The leakage radiation levels outside the unit do not exceed **1 mR/h at a distance of 10 cm** from the nearest accessible surface surrounding the safe with the source in the shielded position.

An HDR suite in this case would have all barriers (walls, floor, and ceiling) of thickness about **18 inches of concrete.**

The American Association of Physicists in Medicine (AAPM) recommends **air kerma strength** (S_k). In practice, S_k is determined from exposure rate measured in free air at a distance of 1 m from the source.

The most suitable method of routine calibration of brachytherapy sources is the **well ionization** or **re-entrant chamber**.

PROSTATE IMPLANTS

Permanent implants with iodine-125 or palladium-103 seeds are used in the treatment of early stage prostate cancer. These are emitters of **characteristic X rays** by electron capture. Dose=125-140 Gy.

Patients are advised not to have prolonged physical contact with pregnant women or young children for a period of 2 months, to abstain from sexual activity for 2 weeks, and to use condoms during intercourse for the first few weeks in case a seed is discharged into the vagina.

Dose delivered:

$$D_{\text{total}} = 1.44\,\dot{D}_0\,T_{1/2}$$

or:

$$\mathrm{D}_{\text{total}} = \dot{D}_0\,T_{\text{av}}$$

INTRAVASCULAR BRACHYTHERAPY

Intraluminal irradiation of coronary and peripheral arteries together with balloon angioplasty and/or stent implantation significantly lowers the rate of neointimal formation, thereby reducing the rate of restenosis to well below 10%

Target volume for intravascular brachytherapy (IVBT) is confined to the region of angioplasty. Typically, it is **2 to 5 cm** in length of artery and 0.5 to **2 mm** in thickness of arterial wall.

β-Particle sources, in general, give higher dose rates and provide greater radiation protection compared to the γ-ray sources.

Intravascular brachytherapy techniques may be classified into two categories: temporary implants (sealed sources or liquid-filled balloons) and permanent implants (radioactive stents). Catheter-based sealed source is the most commonly used method of treatment. It is the preferred method because of its better control of dose delivery.

Typical dosimetric requirements of a temporary intravascular implant are (a) to deliver a target dose of **15 to 20 Gy** to a 2- to 3-cm length of the arterial wall involved at a radial distance of about 2 mm from the source center, (b) to minimize the dose to tissues outside the region of angioplasty, and (c) to take as little time as possible for completion of the procedure, that is, provide target dose rates on the order of 5 Gy/min or greater. These requirements suggest the suitability of high-energy β sources such as **strontium-90, yttrium-90** (Novoste Beta-cath), and **phosphorus-32** (Guidant Galileo) or high-activity γ sources such as **iridium-192** (Cordis Checkmate).

Radionuclide	Half-Life	Photon Energy (MeV)	Half-Value Layer (mm lead)	Exposure Rate Constant (Rcm^2/mCi-h
^{226}Ra	1,600 yr	0.047–2.45 (0.83 avg)	12.0	8.25[a,b] (Rcm^2/mg-h)
^{222}Rn	3.83 days	0.047–2.45 (0.83 avg)	12.0	10.15[a,c]
^{60}Co	5.26 yr	1.17, 1.33	11.0	13.07[c]
^{137}Cs	30.0 yr	0.662	5.5	3.26[c]
^{192}Ir	73.8 days	0.136–1.06 (0.38 avg)	2.5	4.69[c]
^{198}Au	2.7 days	0.412	2.5	2.38[c]
^{125}I	59.4 days	0.028 avg	0.025	1.46[c]
^{103}Pd	17.0 days	0.021 avg	0.008	1.48[c]

Exposure rate constant:

The activity of a radioactive nuclide emitting photons is related to the exposure rate by the exposure rate constant, $\Gamma_{;\delta}$ .In brachytherapy, this constant is usually expressed as numerically equal to the exposure rate in R/h at a point 1 cm from a 1-mCi point source. In the case of radium, the source strength is specified in terms of milligrams of radium instead of mCi.

A new quantity, called **air kerma rate constant**, has been recommended to replace the exposure rate constant.

The SI unit for this quantity is $m^2\ Jkg^{-1}\ h^{-1}\ Ci^{-1}$ or $m^2\ Gy\ Bq^{-1}\ sec^{-1}$.

The **air kerma strength** is defined as the product of air kerma rate in “free space” and the square of the distance of the calibration point from the source center along the perpendicular bisector.

The Sievert integral gives the exposure rate distribution in air and considers only the inverse square law and filtration effects, ignoring attenuation as well as scattering in the surrounding tissue.

The anisotropy factor accounts for the angular dependence of photon absorption and scatter in the encapsulation and the medium.

The radial dose function accounts for radial dependence of photon absorption and scatter in the medium along the transverse axis

Patterson-Parker System:

- In the case of planar implants the uniformity of dose is achieved in parallel planes at 0.5 cm from the implanted plane and within the area bounded by the projection of the peripheral needles on that plane.
- The ratio between the amount of radium in the periphery and the amount of radium over the area itself depends on the size of the implant.

Area	*Fraction Used in Periphery*
<25 cm^2	2/3
25–100 cm^2	½
>100 cm^2	1/3

- The spacing of the needles should not be more than 1 cm from each other or from the crossing ends.
- If the ends of the implant are uncrossed, the effective area of dose uniformity is reduced.[6] The area is, therefore, reduced by 10% for each uncrossed end for table-reading purposes.
- The “stated” dose, determined from the Paterson-Parker tables, is 10% higher than the minimum dose. The maximum dose should not exceed 10% above the stated dose to satisfy the uniformity criterion. The dose is, however, much more nonuniform within the plane of implant. For example, the dose at the surface of the needles is about five times the stated dose.
- In the case of multiple implant planes, the radium should be arranged as in rules 1–3, and the planes should be parallel to each other.

The Paterson-Parker tables are designed to give milligram hours/1,000 roentgens (mg-h/1,000 R) for implants of various sizes, both for planar and volume implants.

To convert Paterson-Parker roentgens to cGy in tissue, one needs to make the following corrections:

(a) Exposure rate constant (Γ): The tables assume Γ = 8.4 Rcm^2/mg-h instead of the current value of 8.25 Rcm^2/mg-h.

(b) A roentgen:cGy factor of 0.957 should be used to convert exposure into dose in muscle.

(c) Oblique filtration: Paterson-Parker tables do not take into account increased attenuation by oblique filtration by the platinum capsule, giving rise to a 2% to 4% error for typical implants.

(d) Paterson-Parker tables are based on exposure in air.

Corrections are needed for tissue attenuation and scattering .

For typical planar and volume implants a combined factor of **0.90** selects an isodose curve approximately equivalent to the Paterson-Parker dosage.

Comparison of Patterson-Parker & Paris systems:

Characteristic	Patterson-Parker	Paris
Linear strength	Constant	Constant
Source distribution	Varies according to size of implant	Uniform
Source spacing	Uniform & constant (1cm)	Uniform but varies according to thickness of tissue
Crossed needles	Used, to increase dose at ends	Not used, hence active length> target length
Dose variation	Less (<=10%)	More (hotter in the middle)

The reference isodose for a Paris implant surrounds the implant within a few millimeters, and its value is approximately equal to 85% of the basal dose, which is defined as the average of the minimum dose between sources.