Chapter 1: Structure of Matter

1.2 The Nucleus
Isotope: same number of protons, different number of neutrons
Isotones: same number of neutrons, different number of protons
Isobars: same number of nucleons, different number of protons
Isomers: same number of protons as well as neutrons

Stable elements in the low atomic number range have an almost equal number of neutrons and protons. However, as the atomic number increases beyond 20, the neutron to proton ratio for stable nuclei becomes greater than 1 and increases with atomic number.

Most stable nuclei have an even number of protons and neutrons (even-even nuclei). This suggests that nuclei gain stability when neutrons and protons are mutually paired.

1.3 Atomic Mass and Energy Units
Mass defect = binding energy of the nucleus.

- some mass is converted to energy to hold the nucleus together. This explains why the mass of an atom is not exactly equal to the sum of the masses of constituent particles

Electron volt (eV): defined as the kinetic energy acquired by an electron in passing through a potential difference of 1 V

Mass of an electron at rest is sometimes expressed in terms of energy equivalent:

- $E_0 = 0.511$ MeV

1.4 Distribution of Orbital Electrons
The maximal number of electrons in an orbit is given by $2n^2$, where n is the orbit number

1.5 Atomic Energy Levels
Binding energy of electrons (energy required to fully remove electron) depends on force of attraction between nucleus and orbital electrons. Binding energies for higher Z atoms are greater because of the greater nuclear charge.

- The further the orbital shell is away from the nucleus, the lesser the binding energies for electrons, and the greater the potential energy for the electrons

1.6 Nuclear Forces
Electrostatic force of attraction binds electrons to nuclei
Strong nuclear force binds protons and neutrons together

1.7 Nuclear Energy Levels 1.8 Particle Radiation 1.9 Electromagnetic Radiation

Miscellaneous:
Radius of an atom = 1×10^{-10} m
Radius of a nucleus = 1×10^{-14} m

Ionization = electron removed from atom
Excitation = electron energy level raised

Number of atoms/gm = Avogadro's # / Atomic Weight
Grams/atom = Atomic Weight / Avogadro's #
Electrons/gm = (Avogadro's # x Z) / Atomic Weight

$E = hv$
$c = v \times \lambda$
$E = (hc)/\lambda$

In order of increasing energy/decreasing wavelength/increasing frequency:

- radio, TV, radar, microwave, infrared, visible light, ultraviolet, x-rays and cosmic rays

1 proton = 1 amu = 931 MeV

Nuclear Transformations

2.1 Radioactivity
2.2 Decay Constant

Exponential equation for radioactive decay: $N = N_0 e^{-\lambda t}$
λ = decay constant (expresses probability that a nucleus will undergo a transition in a stated period of time)
N_0 = initial number of radioactive atoms

2.3 Activity

Activity = rate of decay of a radioactive material

$$A = -\lambda N$$
or
$$A = A_0 e^{-\lambda t}$$

A = activity at time t, and A_0 is original activity
N = number of atoms

Units:

- unit of activity is the curie (Ci)
    - 1 Ci = 3.7×10^{10} Becquerel
- SI unit for activity is Becquerel (Bq)
    - 1 Bq = 1 dps = 2.70×10^{-11} Ci

2.4 The Half-life and the Mean Life

Half life: time required for either the activity or the number of nuclei to decay to half of initial value
$T_{1/2} = 0.693/\lambda$

Mean life: average lifetime for the decay of radioactive atoms

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

Fraction remaining = $(.5)^n$

- where n is the number of half-lives

2.5 Radioactive Series

All elements with Z > 82 (lead) are radioactive. Of the 107 elements known, only the first 92 are naturally occurring.

2.6 Radioactive Equilibrium

If the half life of the parent is longer than that of the daughter, then after a certain time, equilibrium will be achieved (ratio of daughter activity to parent activity is constant)

- Transient equilibrium: if the half-life of the parent is only slightly longer than that of the daughter
- Secular equilibrium: if the half-life of the parent is much longer than that of the daughter

2.7 Modes of Radioactive Decay

A. α Particle Decay
Radioactive nuclides with high atomic numbers (>82) decay most frequently with emission of α particle.

$$X^{A}_{Z} \rightarrow Y^{A-4}_{Z-2} + He^{4}_{2} + Q$$

B. β Particle Decay

Negatron Emission: For radionuclides with excessive number of neutrons.

$$ {}^{A}_{Z}X \rightarrow {}^{A}_{Z+1}Y + {}^{0}_{-1}\beta + Q + \text{antineutrino} $$

Positron Emission: Positron-emitting nuclides have a deficit of neutrons, and their n/p rations are lower than those of stable nuclei. The positron eventually combines with another electron, producing annihilation that results in 2 γ ray photons, each of 0.51 MeV, thus converting two electron masses into energy.

$$ {}^{A}_{Z}X \rightarrow {}^{A}_{Z-1}Y + {}^{0}_{+1}\beta + Q + \text{neutrino} $$

C. Electron Capture

An orbital electron is captured by the nucleus, thus transforming a proton into a neutron. It is an alternative process to positron decay. Electron capture creates an empty hole in the involved shell that is then filled with another outer orbit electron, thus giving rise to characteristic X-rays. There is also emission of Auger electrons, which are monoenergetic electrons produced by the absorption of characteristic X-rays by the atom and reemission of the energy in the form of orbital electrons ejected from the atom.

$$ {}^{A}_{Z}X + {}^{0}_{-1}e \rightarrow {}^{A}_{Z-1}Y + \text{neutrino} + Q $$

D. Internal Conversion

Where excess nuclear energy is passed on to one of the orbital electrons, which is then ejected from the atom.

Internal conversion and electron capture both lead to formation of characteristic x-rays.

2.8 Nuclear Reactions

A. The α, p reaction: α particle interacts with a nucleus to form a compound nucleus, which then disintegrates into a new nucleus by the ejection of a proton.

$$ {}^{A}_{Z}X + {}^{4}_{2}He \rightarrow {}^{A+3}_{Z+1}Y + {}^{1}_{1}H + Q $$

B. The α, n Reaction: bombardment of nucleus with α particle, with subsequent emission of neutron.

C. Proton Bombardment: proton capture by nucleus with emission of a γ ray

D. Deuteron Bombardment: after bombardment, the compound nucleus will emit either a neutron or a proton.

E. Neutron Bombardment: neutron bombardment results in n, γ or n,p reactions The products of these reactions yields β emitters. Slow neutrons or thermal neutrons (neutrons with average energy equal to the energy of thermal agitation in a material, about 0.025 eV at room temp) are effective in producing nuclear transformations.

F. Fission: reaction produced by bombarding certain high atomic number nuclei by neutrons. After absorbing the neutron, the nucleus splits into nuclei of lower atomic number as well as additional neutrons. Typical example is fission of U-235 with slow neutrons.

G. Fusion: Low mass nuclei are combined to produce one nucleus

2.9 Activation of Nuclides
2.10 Nuclear Reactors

Because the neutrons released during fission are fast neutrons, they have to be slowed down to thermal energy (0.025 eV) by collisions with nuclei of low Z material. Such materials are moderators (typically graphite, beryllium, water, and heavy water).

Chapter 3: Production of X-rays

3.1 The X-ray tube: The cathode is the negative electrode, anode is positive electrode.

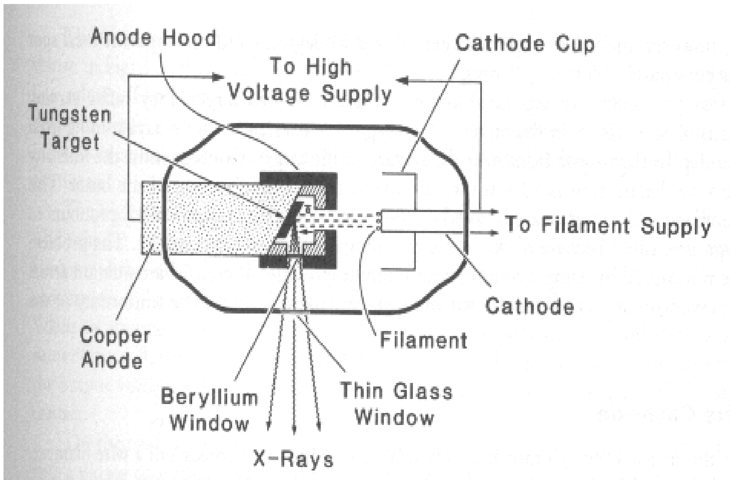


FIG. 3.1. Schematic diagram of a therapy x-ray tube with hooded anode.

Cathode: tungsten filament that when heated, emits electrons ***(thermionic emission)***.

Anode:
- Consists of copper rod, with tungsten target at end of copper rod.
    - Tungsten is an ideal target, due to its high atomic number, and high melting point.
- Efficient removal of heat from the target is an important requirement of the anode. 99% of energy in X-ray production is lost as heat.
- Another important requirement is the optimum size of the target area from which x-rays are emitted ***(focal spot).***
    - The smaller the focal spot, the sharper the radiographic image (important for diagnostic radiology)
        - Smaller focal spots generate more heat per unit area, which can limit currents and exposure
    - Bigger focal spots are okay for therapeutic radiation, since image quality is not a concern
    - The apparent size of the focal spot can be reduced by principle of line focus, where the target is mounted on a steeply inclined surface of the anode
        - Target angles are smaller in diagnostic radiology (to produce smaller focal spots) vs. therapeutic (where having a small focal spot is not as impt)
    - Heel effect: variation of intensity across the x-ray beam.
        - Since x-rays are produced at various depths in the target, they have varying amounts of attenuation in the target.
        - Greater attenuation for x-rays coming from greater depths vs those from near the surface
        - Intensity of the x-ray beam decreases from the cathode to the anode
        - Problem is particularly pronounced in diagnostic tubes because of low x-ray energies and steep target angles
        - Problem can be minimized by using a compensating filter to provide differential attenuation across the beam to compensate for the heel effect and improve uniformity of the beam

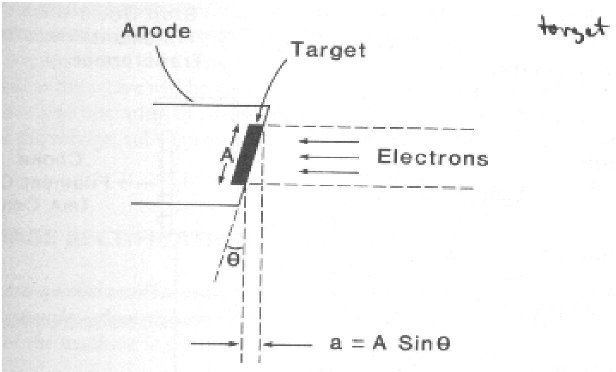


FIG. 3.2. Diagram illustrating the principle of line focus. The side *A* of the actual focal spot is reduced to side *a* of the apparent focal spot. The other dimension (perpendicular to the plane of the paper) of the focal spot remains unchanged.

- When high voltage applied between anode and cathode, the electrons emitted from filament are accelerated towards the anode and strike the target.
- X-rays produced by sudden deflection or acceleration of the electron caused by the attractive force of the tungsten nucleus

3.2 Basic X-ray circuit:

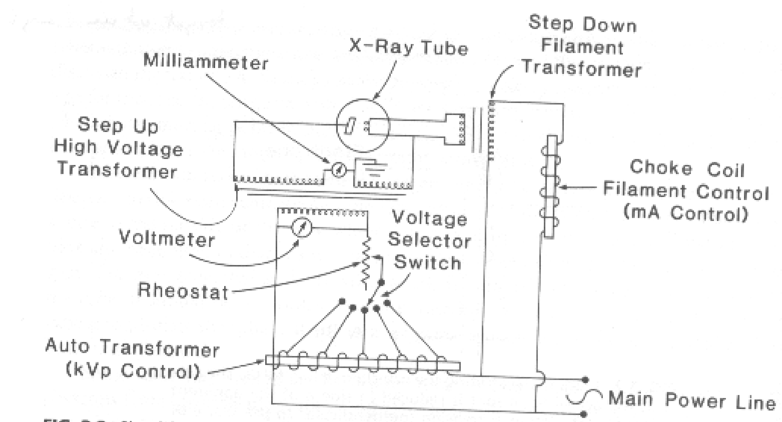


FIG. 3.3. Simplified circuit diagram of a self-rectified x-ray unit.

- The filament supply for electron emission can be accomplished using a step-down transformer in the AC line voltage.
- High voltage to the x-ray tube supplied by the step-up transformer
  - This is connected to an autotransformer and rheostat
    - Autotransformer: provides stepwise adjustment in voltage
    - Rheostat: a variable resister
- Voltage input to the high-tension transformer or the x-ray transformer can be read on a voltmeter in the primary part of its circuit
- Tube current can be read on a milliammeter

Since the anode is positive with respect to the cathode only through half the voltage cycle, the tube current flows through that half of the cycle. During the next half-cycle, the voltage is reversed and current cannot flow in the reverse direction.

- Tube current and x-rays will be generated only in the half of the cycle where the anode is positive, or a ***self-rectified unit***.

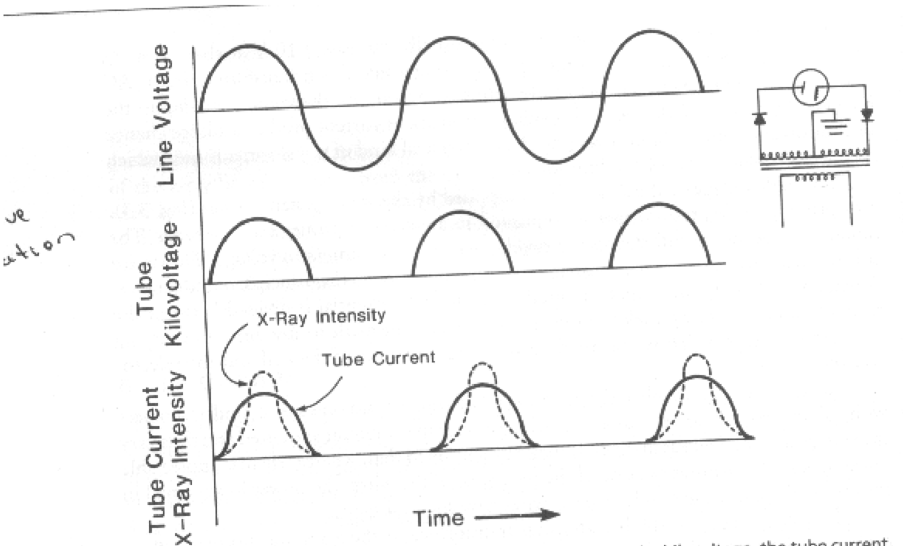


FIG. 3.4. Graphs illustrating the variation with time of the line voltage, the tube kilovoltage, the tube current, and the x-ray intensity for self- or half-wave rectification. The half-wave rectifier circuit is shown on the right. Rectifier indicates the direction of conventional current (opposite to the flow of electrons).

3.3 Voltage rectification: Disadvantage of self-rectified unit is that no x-rays are generated for the half of the cycle of inverse voltage. Tube conduction during inverse voltage can be prevented with placement of rectifiers in series in the high voltage part of the circuit (see above, Figure 3.4)

Rectifiers can also be used to provide ***full-wave rectification***. Anode is positive and cathode negative in both half-cycles of voltage.

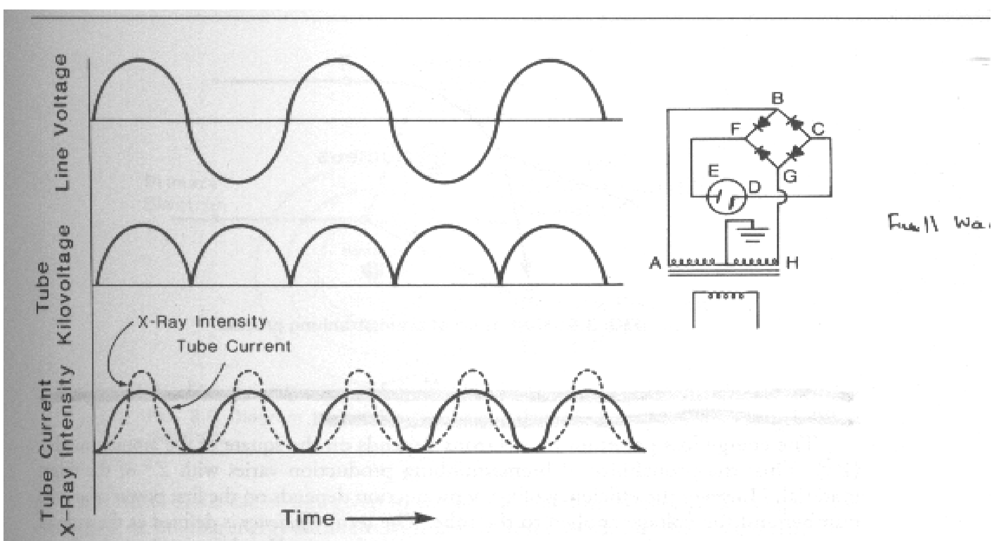


FIG. 3.5. Graphs illustrating the variation with time of the line voltage, the tube kilovoltage, the tube current, and the x-ray intensity for full-wave rectification. The rectifier circuit is shown on the right. The *arrow* symbol on the rectifier diagram indicates the direction of conventional current flow (opposite to the flow of electronic current).

Half-wave has 2 diodes, full-wave has 4 diodes in the form of a bridge.

3.4 Physics of X-ray Production: X-rays can be produced via: 1) Bremsstrahlung or 2) Characteristic x-rays.

<u>*Bremsstrahlung:*</u> "braking radiation". As electrons pass close to a nucleus, it can undergo sudden deflection or acceleration. Part or all of the electron's energy is dissociated from it and propagates as electromagnetic radiation.

- At electron energies < 100 KeV, x-rays are emitted equally in all directions
- As electron energies increase, the x-rays emission becomes increasingly forward

<u>*Characteristic X-rays:*</u> As an electron can interact with atoms of the target by ejecting an orbital electron. Outer shell electrons may then fall down and fill this shell, and in the process, emit characterisitic x-rays.

- the threshold energy an incident electron must possess to eject an electron from an atom of the target is the ***critical absorption energy***

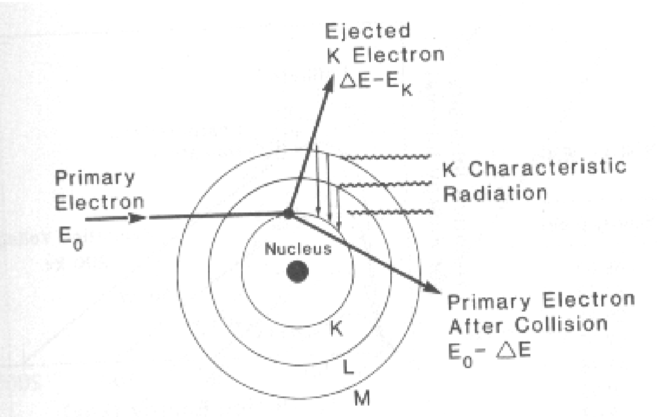


FIG. 3.8. Diagram to explain the production of characteristic radiation.

3.5 X-ray energy spectra:

- Increasing filtration hardens a beam, increasing the average energy by absorbing lower energy components of the spectrum. Though average energy increases, increasing filtration reduces the total intensity of the beam.
- Increasing voltage can also increase the penetrating power of the beam.
- Rule of thumb: average x-ray energy is 1/3 of max energy or kVp

3.6 Operating characteristics

Output of x-ray machine depends on:

- Filament Current
- Tube Current
- Tube Voltage

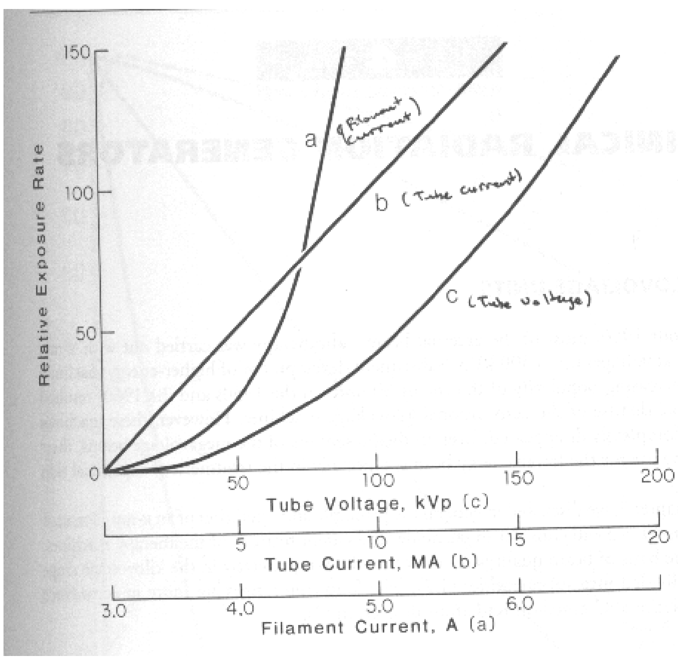


FIG. 3.10. Illustration of typical operating characteristics. Plots of relative exposure rate versus **a**, filament current at a given kVp; **b**, tube current at a given kVp; and **c**, tube voltage at a given tube current.

Chapter 4: Clinical Radiation Generators

4.1 Kilovoltage Units

Grenz-ray Therapy: used to describe treatment with very soft (low-energy) x-rays produced at potentials below 20 kV. No longer used clinically.

Contact Therapy: operates at 40-50 kV. Can be used to treat at very small SSD (2 cm or less). Can produce a very rapidly decreasing depth dose. Useful for tumors no deeper than 1-2 mm. Beam almost completely absorbed within 2 cm of soft tissue.

Superficial Therapy: operates at 50-150 kV potential. SSD typically 15-20 cm. Useful for treating tumors no deeper than 5 mm.

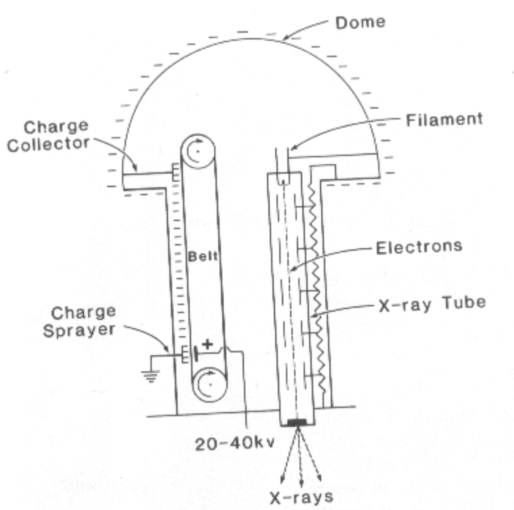
Orthovoltage Therapy (or Deep Therapy): Potential ranges from 150-500 kV. SSD usually set at 50 cm. Max dose at skin surface. 90% IDL at 2 cm depth.

Supervoltage Therapy: Potentials of 500-1000 kV.

Megavoltage Therapy: x-ray beams of energy 1 MV or greater

4.2 Van de Graaf Generator

An electrostatic accelerator that accelerates charged particles to produce high-energy X-rays. In radiotherapy, they usually produce 2 MV x-rays, but are capable of reaching energies of 10 MV.



4.3 Linear Accelerator

Device that uses high-frequency electromagnetic waves to accelerate charged particles (such as electrons) to high energies through a linear tube. Electron beam itself can be used to treat superficial tumors, or it can be used to strike a target to produce x-rays to treat deeper tumors.

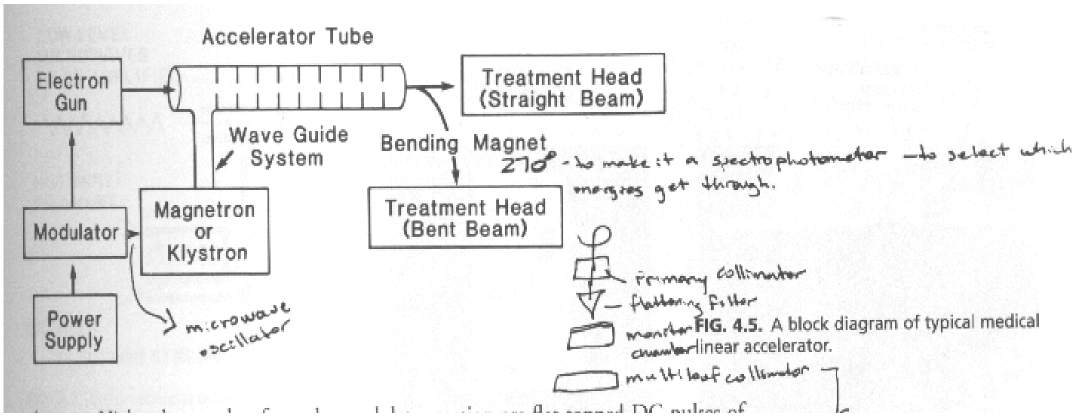


FIG. 4.5. A block diagram of typical medical linear accelerator.

Power supply provides direct current to modulator. High voltage pulses delivered simultaneously to magnetron (or klystron) and electron gun. Microwaves produced in magnetron are pulse injected into accelerator tube via the wave guide. Electrons, produced from electron gun, are also pulse injected into accelerator tube.

In accelerator tube, electrons interact with the electromagnetic field of microwaves, leading to acceleration and gain of energy for the electrons. Once electrons exit, can be used directly to treat superficial tumors, or can strike a target to produce x-rays. For low-energy linacs (up to 6 MV), electrons proceed straight to strike target. For higher energy linacs, a bending magnet is used to bend the electrons to strike the target.

Magnetron: a high power oscillator that produces microwaves

Klystron: Amplifies microwaves, does not produce them. Needs to be driven by a microwave oscillator.

MeV vs. MV: Electron beam designated by MeV (million electron volts) because it is monenergetic. X-ray beam is heterogenous in energy, and designated by MV (megavolts), as if the beam were produced by applying that voltage across an x-ray tube.

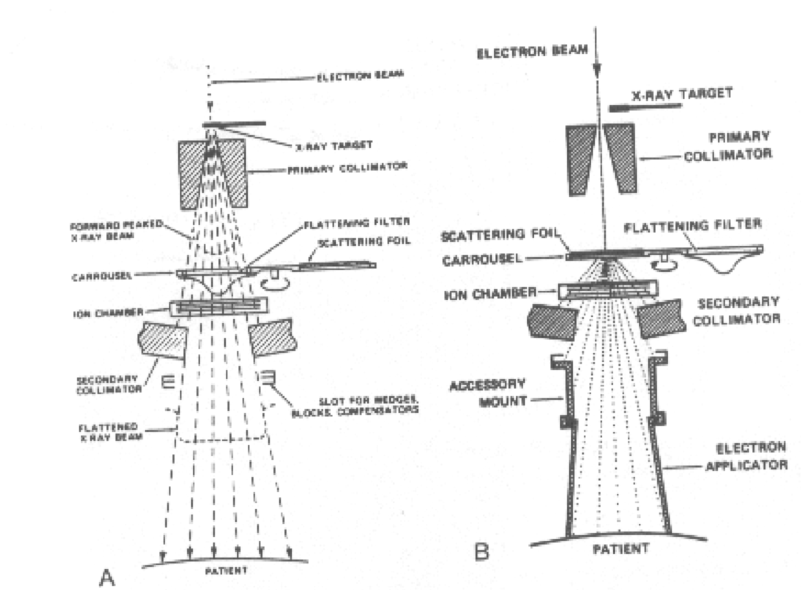


FIG. 4.8. Components of treatment head. A: X-ray therapy mode. B: Electron therapy mode. (From Karzmark CJ, Morton RJ. A primer on theory and operation of linear accelerators in radiation therapy. Rockville, MD: U.S. Department of Health and Human Services, Bureau of Radiological Health, 1981, with permission.)

The Electron Beam: as electron beam exits accelerator tube, its diameter is 3 mm. For electron therapy, electron beam strikes a scattering foil (usually thin layer of lead), to spread the beam and get uniform electron fluence across the treatment field. Most electrons are scattered instead of undergoing bremsstrahlung, although a small fraction will undergo bremsstrahlung, and appear as x-ray contamination of the electron beam.

The Treatment Head: Consists of a thick shell of high-density shielding material, which shields against leakage radiation. The treatment head contains an x-ray target, scattering foil, flattening filter, ion chamber, fixed and movable collimator, and light localizer system.

Flattening Filter: usually made of lead. In the production of x-rays, used to make beam intensity uniform across the field.

Beam Collimation and Monitoring: Two collimators used (beam first passes through a fixed, primary collimator, before passing through a second, movable collimator). Movable collimators provide opening from 0 x 0 to maximum field size of 40 x 40 cm at 100 cm from x-ray source. The field size definition is provided by a light localizing system in the treatment head.

Electrons scatter readily in air, so beam collimation must be achieved close to skin surface. Auxiliary collimators for electrons in the form of trimmers (attachable cones) extended down to skin surface accomplishes this.

4.4 Betatron: Electrons introduced to doughnut tube via injector. Electrons are accelerated in presence of magnetic field, and once they reach maximum energy, they either: a) strike a target to produce x-rays via bremsstrahlung or b) strike a scattering foil to produce electrons.

X-ray dose rate capabilities are low compared to linacs. In electron therapy mode, the dose rate is adequate. Not frequently used anymore because linacs have capability to deliver higher dose rate electrons and x-rays.

4.5 Microtron:

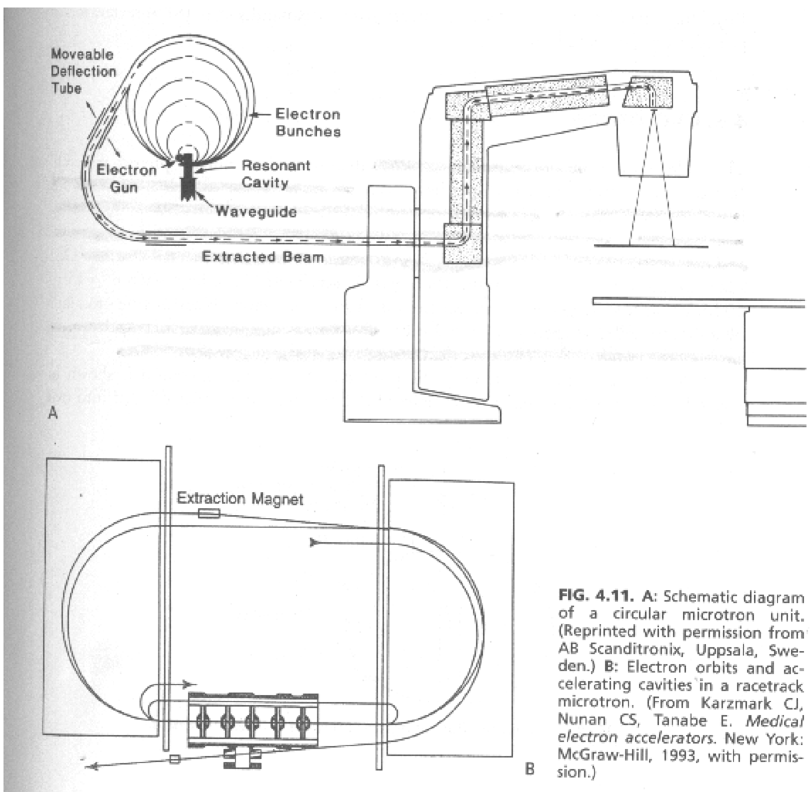


FIG. 4.11. **A**: Schematic diagram of a circular microtron unit. (Reprinted with permission from AB Scanditronix, Uppsala, Sweden.) **B**: Electron orbits and accelerating cavities in a racetrack microtron. (From Karzmark CJ, Nunan CS, Tanabe E. *Medical electron accelerators*. New York: McGraw-Hill, 1993, with permission.)

Purpose of the deflection tube is to extract electrons at the appropriate time. As the electrons receive higher and higher energy by repeated passes through the cavity, the describe orbits of increasing radius in the magnetic field.

Advantages vs. linacs: simplicity, easy energy selection, small beam energy spread, smaller size.

4.6 Cyclotron: a charged particle accelerator. Used as a source of high-energy protons, and also for generating neutron beams (in which case deuterons are accelerated to high energies, and strike a low atomic number target (beryllium) to produce neutrons). Also used as a particle accelerator for the production of some radionuclides.

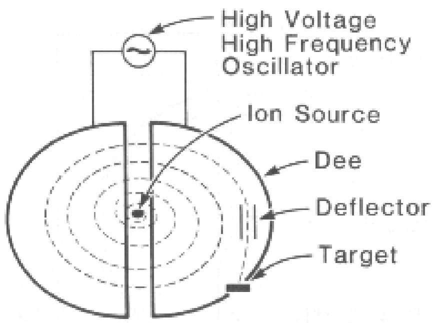


FIG. 4.12. Diagram illustrating the principle of operation of a cyclotron.

Positively charged particles are injected into the chamber in between the two "D's". These particles travel in circular orbits and increase energy as the radius of their orbit increase. As the particle reaches high velocity, further acceleration can cause an increase in mass, which can ultimately lead to a decrease in particle velocity.

4.7 Machines using radionuclides: Radium-226, Cesium-137, and Cobalt-60 have been used as sources of γ rays for teletherapy (term applied to external beam treatments in which the source of radiation is at a large distance from the patient). Cobalt-60 has proved to be the most suitable for external beam radiotherapy due to higher specific activity, greater radiation output per curie, and higher average photon energy.

TABLE 4.1. TELETHERAPY SOURCE CHARACTERISTICS

Radionuclide	Half-Life (yr)	γ-Ray Energy MeV	Γ Value[a] $\left(\frac{Rm^2}{Ci-h}\right)$	Specific Activity Achieved in Practice (Ci/g)
Radium-226 (filtered by 0.5 mm Pt)	1,622	0.83 (avg.)	0.825	~0.98
Cesium-137	30.0	0.66	0.326	~50
Cobalt-60	5.26	1.17, 1.33	1.30	~200

[a] Exposure rate constant (Γ) is discussed in Chapter 8. The higher the Γ value, the greater will be the exposure rate or output per curie of the teletherapy source.

Cobalt-60 Unit: Source is a cylinder of diameter 1-2 cm. The fact that source is not a point source gives rise to geometric penumbra.

- Penumbra: region at the edge of a radiation beam over which the dose rate changes rapidly as a function of distance from the beam axis
  - Transmission penumbra: region irradiated by photons which are transmitted through the edge of the collimator block

The geometric penumbra, at depth d, can be calculated by the following equation:

$P_d = s(SSD + d - SDD)/ SDD$

s=source diameter, SSD = source to surface distance, SDD = source to diaphragm distance

To reduce penumbra: 2 options

1) Penumbra trimmers: effectively increases the SDD, thereby decreasing geometric penumbra
2) Placing secondary blocks close to the patient, to redefine or shape the field. Blocks should not be placed closer than 15-20 cm from pt b/c of excessive electron contaminants produced by the block carrying tray.

At depth in a 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 pt.

- Physical penumbra width: defined as the lateral distance between two specified isodose curves at a specified depth

4.8 Heavy Particle Beams: offer advantages wrt dose localization and therapeutic gain, but their use is still limited clinically.

Neutrons: Produced by deuterium-tritium generators, cyclotrons, or linacs. The bombarding particles are usually deuterons or protons, and target is usually beryllium, except in the deuterium-tritium generator, where tritium is the target.

Protons and Heavy Ions: Proton beams for therapy range in energy from 150 to 250 MeV, produced by cyclotrons or linacs. Major advantage of protons and heavy ions is that as beam traverses tissues, dose deposited is constant with depth until the end of the range where the dose peaks out to a high value followed by a rapid falloff to zero (Bragg peak).

Negative pions: Protons and neutrons in the nucleus are held together by a mutual exchange of pions. Negative pions have been used for RT. Produced when protons strike a beryllium target. Negative pions are extracted from the target using bending and focusing magnets. Negative pions of 100 MeV are of interest in RT, providing range in water of 24 cm. Bragg peak exhibited by negative pions is more pronounced than other heavy particles.

Chapter 5: Interactions of Ionizing Radiation

5.1 Ionization

Charged particles are directly ionizing, as they produce ionization by collision as they penetrate matter. The energy of the incident particle is lost in a large number of small increments along the ionization track in the medium, with an occasional interaction in which the ejected electron receives sufficient energy to produce a secondary track of its own, known as a delta ray.

Uncharged particles are indirectly ionizing, and liberate directly ionizing particles from matter when they interact with matter.

Ionizing photons interact with atoms of a material to produce high-speed electrons by three major processes: photoelectric effect, Compton effect, and pair production. 2 other processes of interaction are coherent scattering and photo disintegration (photo disintegration occurs between photons and nucleus, and is important only at very high photon energies (> 10 MeV)).

5.3 Photon Beam Attenuation

$I(x) = I_o e^{-\mu x}$

- $I(x)$ is the intensity transmitted by a thickness x and I_o is the incident intensity. μ is the attenuation coefficient.
- Exponential attenuation applies strictly to a monoenergetic beam

HVL = half value layer

- thickness of an absorber required to attenuate the intensity of the beam to half its original value.
- HVL = $0.693/\mu$
- For a heterogeneous beam, the first HVL is less than subsequent HVL's.
    - Due to beam hardening as beam passes through each HVL

5.6 Coherent Scattering: Incident photon interacts with an oscillating electron, and is scattered, maintaining the same wavelength. No energy is transferred. Only important for very low photon energy (< 10 keV) and high Z materials. This phenomenon is only of academic interest in radiation therapy.

5.7 Photoelectric Effect: Most probable when energy of incident photon is equal to or slightly greater than the binding energy of the electron. A photon interacts with an atom, the entire energy of the photon is absorbed by the atom, and transferred to an orbital electron which is then ejected. The ejected electron creates a vacancy within its shell, which can then be filled by an electron from an outer shell, resulting in the emission of characteristic x-rays. Alternatively, these characteristic x-rays can be absorbed by the atom to produce the emission of Auger electrons (monoenergetic electrons produced by the abosorption of characteristic x-rays internally by the atom).

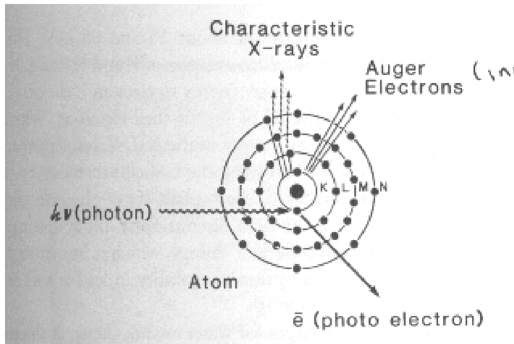


FIG. 5.5. Illustration of the photoelectric effect.

Depends strongly on the atomic number. The mass photoelectric attenuation coefficient is $\propto Z^3/E^3$

5.8 Compton Effect: process where photon interacts with an atomic electron as if it were a "free electron" (binding energy of electron is much less than that of the bombarding photon). The electron receives some energy from the photon and is emitted. The photon is scattered at reduced energy. Unlike photoelectric effect, compton effect is dependent not on atomic number, but on number of electrons per gram. Although the number of electrons/gram decreases slowly with atomic number, most materials except hydrogen can be considered as having the same number of electrons/gram. Hydrogen has double the number of electrons/gram.

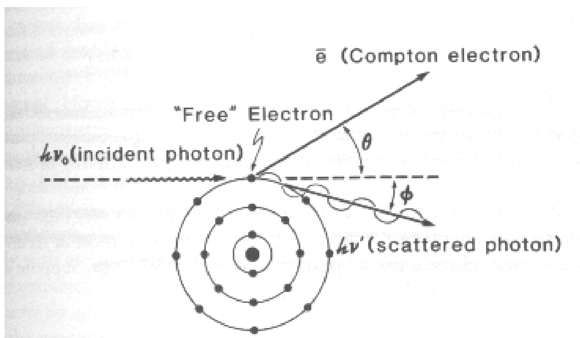


FIG. 5.7. Diagram illustrating the Compton effect.

Special cases of Compton Effect:

1) Direct hit: where photon makes a direct hit with electron. Electron will travel forward (at angle 0 degrees), and photon will travel backward (at angle 180 degrees).
2) Grazing Hit: Electron emitted at right angle and scattered photon will go in forward direction (0 degrees). No energy lost for the photon.
3) Photon scattered at right angle to original direction.

5.9 Pair Production: Threshold energy of this process is 1.02 MeV. Photon interacts strongly with the electromagnetic field of the nucleus and gives up all of its energy to produce a positive electron (positron) and negative electron. Photon energy in excess of the threshold is shared by the electrons as kinetic energy.

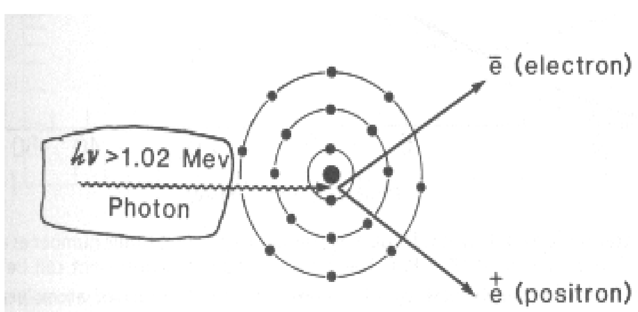


FIG. 5.9. Diagram illustrating the pair production process.

The positron will eventually combine with a free electron to give rise to two annihilation photons, each having 0.51 MeV energy. These two photons are ejected in opposite directions.

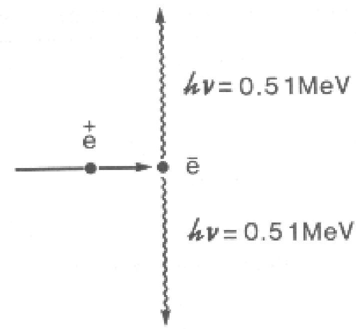


FIG. 5.10. Diagram illustrating the production of annihilation radiation.

Probability of this interaction increases with atomic number.

5.10 Relative Importance of Various Types of Interactions:

TABLE 5.2. RELATIVE IMPORTANCE OF PHOTOELECTRIC (τ), COMPTON (σ), AND PAIR PRODUCTION (Π) PROCESSES IN WATER

Photon Energy (MeV)	Relative Number of Interactions (%)		
	τ	σ	Π
0.01	95	5	0
0.026	50	50	0
0.060	7	93	0
0.150	0	100	0
4.00	0	94	6
10.00	0	77	23
24.00	0	50	50
100.00	0	16	84

Data from Johns HE, Cunningham JR. The physics of radiology. 3rd ed. Springfield, IL: Charles C Thomas, 1969.

5.11 Interactions of Charged Particles: Whereas photons interact with matter by photoelectric, Compton, or pair production, charged particles interact via ionization and excitation. This is mediated by Coulomb force between the traveling particle and orbital electrons (which results in ionization and excitation of atoms) or between the traveling particle and the nucleus of atoms (resulting in bremsstrahlung).

Stopping power = rate of kinetic energy loss per unit path length of the particle

- proportional to the square of the particle charge, and inversely proportional to the square of the velocity
    - As particle slows down, rate of its energy loss increases, and so does ionization or absorbed dose of the medium
    - This peaking of dose near the end of particle range is known as the Bragg Peak
        - Gives proton and heavy charged particle ability to concentrate dose inside the target volume and minimize dose to surrounding normal tissue

5.12 Interactions of Neutrons: Neutrons interact by two processes

1) recoiling protons from hydrogen and recoiling heavy nuclei from other elements
2) nuclear disintegrations

Dose deposited in tissue from a high-energy neutron beam is predominantly contributed by recoil protons. Nuclear disintegrations produced by neutrons result in emission of heavy charged particles, neutrons, and gamma rays and give rise to about 30% of tissue dose.

Why is hydrogenous materials such as parrafin wax or polyethylene good absorbers of neutrons, while lead is not?

- Because when neutrons collide with particles of equal mass, energy transfer is very efficient. When neutrons collide with heavier nuclei, the neutron loses very little energy

Measurement of Ionizing Radiation

6.2 The Roentgen

The roentgen is unit of exposure. Exposure is a measure of ionization produced in air by photons. Exposure can also be defined as coulombs per kilogram air.

$$X = dQ/dm$$

The definition of roentgen is satisfied under condition of electronic equilibrium, where the electrons produced in the specified volume for measurement that deposit their energy outside the region of ion collection and are not measured are equal to the electrons produced outside the specified volume that enter the ion-collecting region and produce ionization and are measured.

6.3 Free-air Ionization Chamber

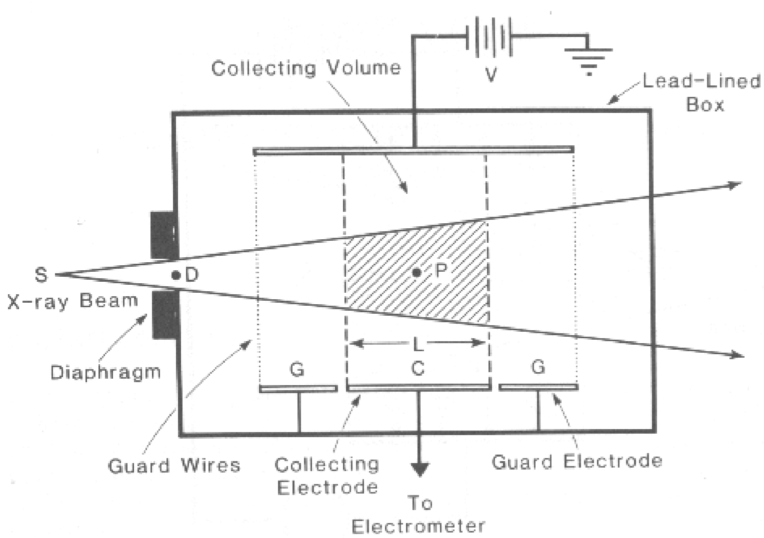


FIG. 6.2. A schematic diagram of a free-air chamber.

An x-ray beam, originating from focal spot S, is defined by diaphragm D, and passes centrally between a pair of parallel plates. A high-voltage is applied between the plates to collect ions produced in the air between the plates. The ionization is measured for a length L defined by the limiting lines of force to the edges of the collection plate C.

Limitations: high-energy x-ray beams. As photon energy increases, the range of the electrons liberated in air increases. 3 MeV is the upper limit photon energy above which the roentgen cannot be accurately measured.

Too delicate and bulky for routine use. Main function is in standardizing laboratories where they can be used to calibrate field instruments, such as thimble chambers.

6.4 Thimble Chambers

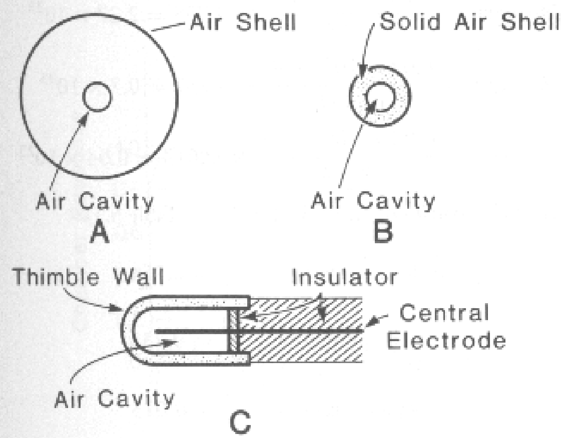


FIG. 6.3. Schematic diagram illustrating the nature of the thimble ionization chamber. **A:** Air shell with air cavity. **B:** Solid air shell with air cavity. **C:** The thimble chamber.

Thimble wall forms one electrode. Other electrode is a rod of low atomic number. Voltage applied between the two electrodes to collect 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.

Desirable Chamber Characteristics

1. Minimal variation in sensitivity or exposure calibration factor over a wide range of photon energies
2. Suitable volume to allow measurements for the expected range of exposures. The sensitivity (charge measured per roentgen) is directly proportional to the chamber sensitive volume.
3. Minimal variation in sensitivity with direction of incident radiation.
4. Minimal stem "leakage". A chamber is known to have stem leakage if it records ionization produced anywhere other than its sensitive volume.
5. Chamber should have been calibrated for exposure against a standard instrument for all radiation qualities of interest
6. Should be minimal ion recombination losses.

6.5 Practical Thimble Chambers

Condenser Chamber: a thimble ionization chamber connected to a condenser. Suitable for measuring exposure rate in air for low-energy beams ($\leq$ 2 MeV). For higher energy radiation, design of the stem and stem leakage create dosimetric problems.

- Sensitivity of a chamber = voltage drop per roentgen
    - Chamber sensitivity directly proportional to the chamber volume and inversely proportional to the chamber capacitance
- Stem Effect: if the irradiation of the stem gives rise to ionization that can be measured by the chamber, the chamber reading depends on the amt of the stem in the beam. Thus a stem correction will be needed whenever the length of the stem irradiated differs from that irradiated at the time of chamber calibration.

Farmer Chamber: chamber designed that provided a stable and reliable secondary standard for x-rays and gamma rays for all energies in the therapeutic range.

For lower energy radiation (superficial or orthovoltage ranges), thimble chambers are calibrated and used without a build-up cap. For higher energies, a buildup cap made of Lucite is used unless the chamber wall is already thick enough to produce electronic equilibrium.

6.6 Electrometers

String Electrometer: basically a charge measuring device. Commonly used for measurement of charge on a condenser chamber.

6.7 Special Chambers

Cylindrical thimble chamber most often used for exposure calibration of radiation beams when the dose gradient across the chamber volume is minimal. Not suitable for surface dose measurements (due to dose buildup effect at in first few millimeters at surface).

Extrapolation chambers: ionization chamber designed for measuring surface dose. Used for measurement of dose in the superficial layers of a medium and the dosimetry of electrons and beta-particles

Parallel-plate Chambers: Also used for measurements of surface dose.

6.8 Ion Collection

Saturation: As the voltage difference between electrodes of an ion chamber exposed to radiation is increased, the ionization current increases at first almost linearly, and later more slowly. The curve finally approaches a saturation value for the given exposure rate

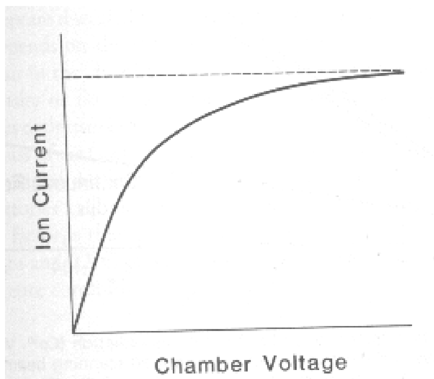


FIG. 6.16. Saturation curve for an ion chamber.

If voltage is increased much beyond saturation, the ions can gain enough energy (accelerated by the electric field) to produce ionization by collision with gas molecules. This results in a rapid multiplication of ions, and the current again becomes strongly dependent on the voltage.

The chamber should be used in the saturation region so that small changes in voltage do not result in changes in the ionic current.

Collection Efficiency: defined as the ratio of the number of ions collected to the number produced. Especially at high ionization intensity, significant loss of charge by recombination may occur (even at maximum possible chamber voltages). Under these conditions, recombination losses may have to be accepted and correction applied for these losses. When possible, voltage on the chamber should be arranged to give less than 1% loss of charge by recombination (collection efficiency of better than 99%).

6.9 Chamber Polarity Effects

Sometimes, for a given exposure the ionic charge collected by an ion chamber changes in magnitude as the polarity of the collecting voltage is reversed. Worse for electrons vs. photons. Also, effect increases with decreasing electron energy. Polarity effect very much dependent on chamber design and irradiation conditions.

6.11 Measurement of Exposure

Exposure in units of roentgen can be measured with following equation:

$X = M \times N_c \times C_{T,P} \times C_s \times C_{st}$

X= exposure that would be expected in free air at the point of measurement in the absence of the chamber
M = reading obtained for a for a given exposure

N_c = exposure calibration factor
$C_{T,P}$ = correction for temp and pressure = (760/P) x (273.2 + t)/295.2

- P in mmHG, T in Celsius

C_s = correction for loss of ionization as a result of recombination
C_{st} = stem leakage correction

Khan: Chapter 7
Quality of X-ray Beams

Half Value Layer (HVL): thickness of an absorber of specified composition required to attenuate the intensity of the beam to half its original value.

- a term to describe the ***quality*** of an x-ray beam: the penetrating ability of a beam
- for beams in superficial and orthovoltage range: quality described by kVp and HVL
- for megavoltage x-rays, the quality is specified by the peak energy, and rarely by the HVL
- HVL = 0.693/μ
    - μ = linear attenuation coefficient

Determination of x-ray tube potential: difficult b/c not easily accessible for direct voltage measurement. Indirect methods often used to measure kVp. However, if access can be achieved, direct measurements can be made by ***precision voltage dividers*** or a ***sphere-gap apparatus.***

Filters: increasing filter thickness leads to increasing average energy from beam hardening, and decreased intensity

Energy	Filter
Diagnostic and Superficial x-ray	Aluminum: used to harden beam
Orthovoltage range	Combination filters are used • often contain tin, copper, and aluminum • aka ***Thoraeus filters*** • highest atomic number material placed closest to x-ray target
Cesium and Cobalt teletherapy units	Beams are monoenergetic. Filter not needed.
Megavoltage	Beam hardened by inherent filtration of transmission target and by transmission through flattening filter • primary purpose of flattening filter is to make beam intensity uniform in cross section rather than improve beam quality

Effective Energy of an x-ray beam: the energy in a monoenergetic beam which is attenuated at the same rate as the radiation in question.

- determined by finding the energy of monoenergetic photons which have the same μ as the given beam.

Measurement of Megavoltage Beam Energy: Most practical method of determining megavoltage beam energy is by measuring PDD distribution, TAR, or TMR and comparing them with the published data.

Miscellaneous:

- Average beam energy = Emax/3
- Although lead is commonly used to express HVL for megavoltage beams, it may not be the best choice for characterizing beam quality in this energy range. Low atomic number materials such as water are more sensitive to changes in spectral quality of megavoltage x-rays

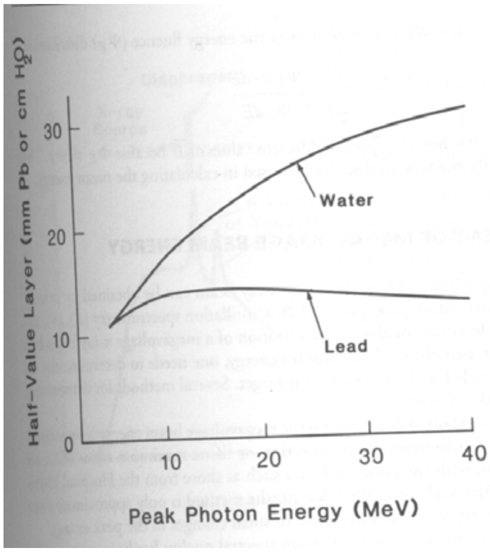


FIG. 7.5. Half-value layer as a function of peak photon energy for water and lead. *Note:* Since these data were calculated from thin-target Schiff (12) spectra, HVL values plotted here are slightly lower than those measured in practical radiotherapy machines. (Data from Nath R, Schulz RJ. On the choice of material for half-value-layer measurements for megavoltage x-rays. *Med Phys* 1977;4:132, with permission.)

Remaining questions:

- How can effective photon energy be increased? Increased tube voltage, increased tube current, decreased filtration?
- How to increase bony resolution in a simulator? Use of kVp and mA

Measurement of Energy Spectrum: HVL cannot be used in systems that are sensitive to spectral distribution of photons. In such instances, spectrometry can be used, including ***scintillation spectrometry***

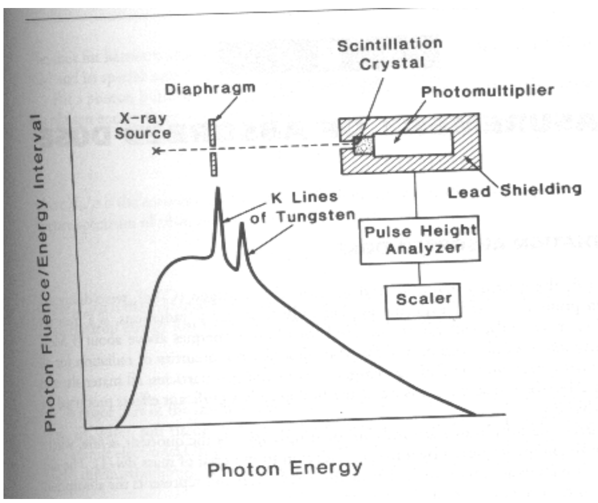


FIG. 7.6. Energy spectrum of an x-ray beam determined by scintillation spectrometer (shown in the inset).

Chapter 8: Measurement of Absorbed Dose

8.1 Radiation Absorbed Dose: Exposure (units of roentgen, or 1 R = 2.58×10^{-4} C/kg), applies only to x- and gamma radiations, is a measure of ionization in air only, and cannot be used for photon energies > 3 MeV.

Absorbed dose defined to described the quantity of radiation for all types of ionizing radiation, in all materials, and for all energies.

- Old unit = rad
    - 1 rad = 100 erg/gm
- New unit = Gy
- 1 Gy = 100 rad
- 1 Sv = 100 rem

8.2 Relationship Between Kerma, Exposure, and Absorbed Dose

Kerma (kinetic energy released in the medium).

- Units of J/kg. Also has unit of rad.
- SI unit is Gy

For a photon beam traversing a medium, the kerma at a point is directly proportional to the photon energy fluence. A major part of the initial kinetic energy of electrons in low Z materials is expended by inelastic collisions (ionization and excitation) with atomic electrons. Only a small part is expended in radiation collisions with atomic nuclei (bremsstrahlung, < 1% of radiation). Kerma can be divided into two parts, the collision and the radiation parts of kerma.

$$K = K_{col} + K_{rad}$$

Under transient charged particle equilibrium, dose = Kcol x constant (close to 1)

Miscellaneous: energy required to produce ion pair in air = 33.97 eV/ion pair

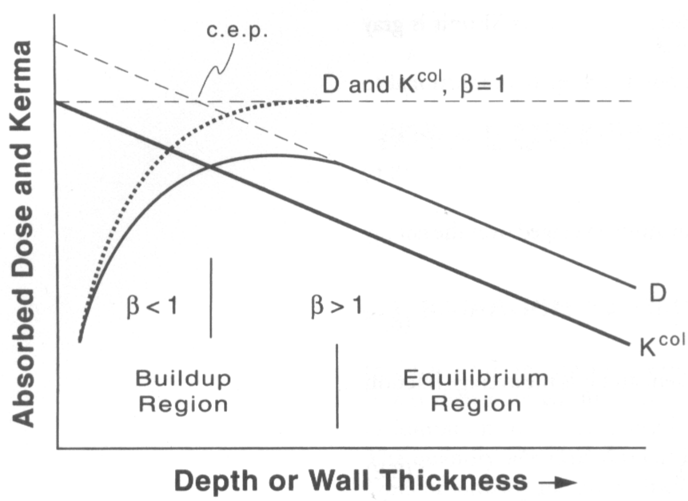


FIG. 8.1. Relationship between absorbed dose (*D*) and collision kerma (K^{col}) for a megavoltage photon beam. *β* is the ratio of absorbed dose to collision kerma. The point designated as c.e.p. is the center of electron production (see text). (From Loevinger R. A formalism for calculation of absorbed dose to a medium from photon and electron beams. *Med Phys* 1981;8:1, with permission.)

8.3 Calculation of Absorbed Dose from Exposure:

A. *Absorbed Dose to Air:* limited to photon energies up to Co-60.

$$D_{air}(rad) = 0.876(rad/R) \times X(R)$$

- 0.876 is the roentgen to rad conversion factor for air

B. *Abosrbed Dose to Any Medium:*

$$D_{med} = f \text{ factor} \times X \times A$$

- A = transmission factor
- f factor: roentgen to rad conversion factor
    - Function of the medium composition as well as the photon energy
    - In the Compton range of energies, the f factor varies as a function of the number of electrons per gram. Since the number of electrons per gram for bone is slightly less than for air, water, or fat, the f factor for bone is also slightly lower than for air and water in the Compton region of the megavoltage energies. F factor is not defined beyond 3 MeV since the roentgen is not defined beyond this energy

***C.* Dose Calibration with Ion Chamber in Air**

D. *Dose Measurement from Exposure with Ion Chamber in a Medium*

8.4 The Bragg-Gray Cavity Theory

Can be used to calculate dose directly from ion chamber measurements in a medium without the restrictions associated with exposure

- Limitations with exposure: cannot be used for energies > 3 MeV, limited to x- and gamma radiation, requires electronic equilibrium

According to the Bragg-Gray Cavity 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 Bragg-Gray Cavity Relationship is satisfied.

$$D_{\text{med}} = J_g \cdot \frac{\overline{W}}{e} \cdot (\overline{S}/\rho)_g^{\text{med}} \qquad (8.30)$$

where D_{med} is the absorbed dose in the medium (in the absence of the cavity), J_g is the ionization charge of one sign produced per unit mass of the cavity gas, and $(\overline{S}/\rho)_g^{\text{med}}$ is a weighted mean ratio of the mass stopping power of the medium to that of the gas for the electrons crossing the cavity. The product of $J_g\left(\frac{\overline{W}}{e}\right)$ is the energy absorbed per unit mass of the cavity gas.

A. *Stopping Power:* Refers to the energy loss by electrons per unit path length of a material.

Spencer-Attix Formulation: a good approximation for the stopping power ratio for an air-filled cavity in a medium such as water under electron irradiation.

B) Chamber Volume

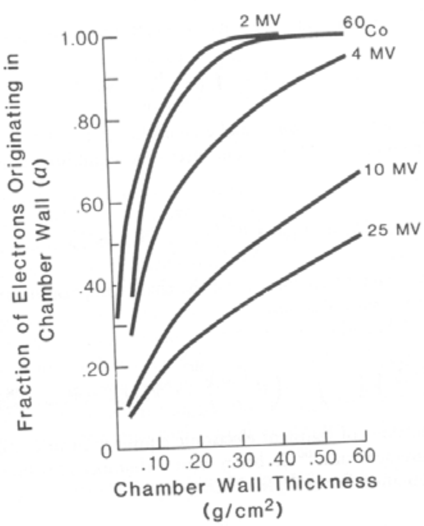


FIG. 8.4. The fraction, α, of cavity ionization due to electrons generated in the chamber wall, plotted as a function of wall thickness. (From Lempert GD, Nath R, Schulz RJ. Fraction of ionization from electrons arising in the wall of an ionization chamber. *Med Phys* 1983;10:1, with permission.)

This fraction is independent of wall composition or build-up cap, as long as it is composed of low Z material

C) Effective Point of Measurement

For plane parallel chambers: if the chamber has a small plate separation, and the electron fluence is mostly forward directed, it is reasonable to assume that the point of measurement is the front surface of the cavity.

For cylindrical chambers: electrons traversing a cylindrical chamber will enter the chamber at different distances from the center of the chamber. The theoretic point of measurement for a cylindrical chamber in a unidirectional beam is displaced by 0.85r from the center and toward the source.

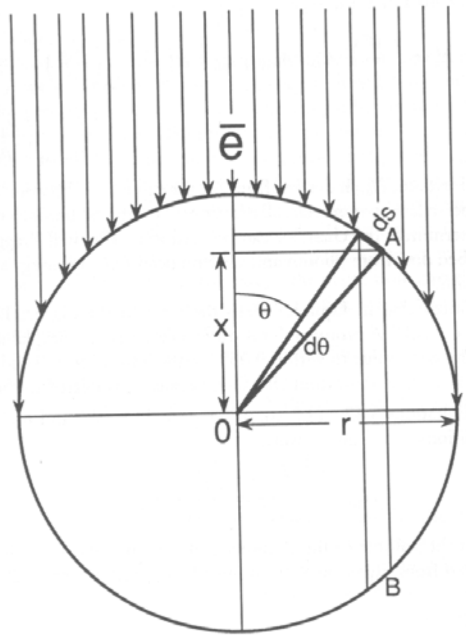


FIG. 8.5. Diagram to illustrate the determination of effective point of measurement for a cylindrical chamber exposed to a unidirectional electron beam.

8.6 AAPM TG-51 Protocol

The basic TG-51 equation for absorbed calibration is as follows:

$$D_w^Q = M k_Q N_{D,w}^{60_{Co}} \tag{8.76}$$

where D_w^Q is the absorbed dose to water at the reference point of measurement in a beam of quality Q; M is the electrometer reading that has been fully corrected for ion recombination, environmental temperature and pressure, electrometer calibration and chamber polarity effects; k_Q is the quality conversion factor that converts the absorbed-dose-to-water calibration factor for a ^{60}Co beam into the calibration factor for an arbitrary beam of quality Q; and $N_{D,w}^{60_{Co}}$ is the absorbed-dose-to-water calibration factor for the chamber in a ^{60}Co beam under reference conditions.

Beam Quality, Q:

Photon beams:

- %dd(10)x = photon component of the photon beam percentage depth dose at 10-cm depth in a 10 x 10 cm field on the surface of a water phantom at an SSD of 100 cm
- electron contamination eliminated by placing 1-mm lead foil at 50 cm from phantom surface for energies > 10 MV

Electron beams:

- beam quality for e- beam dosimetry is specified by R50
  - R50 = depth in water at which PDD is 50% for a broad beam, at SSD 100 cm

*Calibration Phantoms:*TG-51 requires that calibration of beams be performed in a water phantom, at least 30 x 30 x 30 cm

Charge Measurement:

The charge reading, M is = M(raw) x P(ion) x P(T,P) x P(elec) x P(pol)

- M(raw) = raw chamber reading
- P(ion) = ion recombination correction
- P(T,P) = air temperature and pressure correction

- P(elec) = electrometer correction factor
- P(pol) = polarity correction

$P_{T,P}$= (760/P) x ((273.2 + T)/295)   for P in mmHg, and T in Celsius

8.9 Other Methods of Measuring Absorbed Dose

A. Calorimetry: energy absorbed ultimately appears as heat energy. Increase in temp can be measured and related to the energy absorbed.

- To measure such small T rises, thermistors are used
  - Thermistors: semiconductors which show large change in electrical resistance with a small change in T.
  - By measuring change in resistance by apparatus such as a Wheatstone Bridge, can calculate absorbed dose

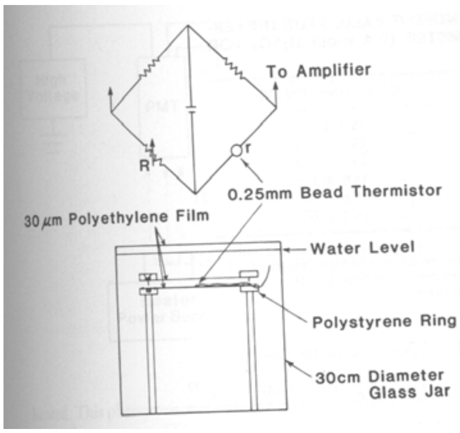


FIG. 8.9. Schematic diagram of Domen's calorimeter. (Redrawn from Domen SR. Absorbed dose water calorimeter. *Med Phys* 1980;7:157.)

B. Chemical Dosimetry: energy absorbed can produce chemical change, and determination of this change can be used to measure absorbed dose
   a. Ferrous Sulfate (Fricke) Dosimeter: When irradiated, ferrous ions, Fe2+, are oxidized by radiation to ferric ions, Fe3+. Ferric ion concenctration determined by spectrophotometry.
      i. G Value: number of molecules produced per 100 eV of energy absorbed. Once yield of ferric ions known, can use the G Value to determine the energy absorbed.

C. Solid State Methods: most widely used are TLD's, diodes, and film
   a. TLDs (Thermoluminescence Dosimetry): when crystal is irradiated, fraction of the absorbed energy is stored and can be recovered later as visible light when the material is heated
      i. Irradiated material is heated, emitted light is measured by a photomultiplier tube which converts light into electrical current, and the current is then amplified and measured by a recorder or counter.
      ii. LiF is is most frequently used material
      iii. Because most phosphors contain many traps at various energy levels in the forbidden band, the glow curve may consist of a number of glow peaks, which correspond to different trapped energy levels.

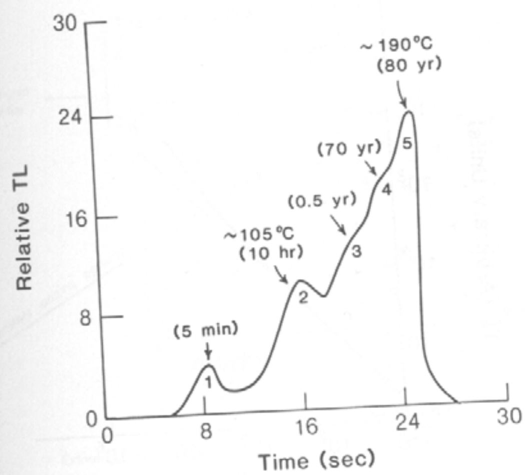


FIG. 8.12. An example of glow curve of LiF (TLD-100) after phosphor has been annealed at 400°C for 1 h and read immediately after irradiation to 100 R. (From Zimmerman DW, Rhyner CR, Cameron JR. Thermal annealing effects on thermoluminescence of LiF. *Health Phys* 1966;12:525, with permission.)

iv. Annealing: TLD material must be annealed to remove residual effects
   1. Pre-RT annealing: LiF heated 1 hr at 400 C, then 24 hr at 80 C. This removes peaks 1 and 2 of the glow curve, which makes the glow curve more stable and predictable
   2. Post-RT annealing can also be done to remove peaks 1 and 2: 10 minutes at 100 C

v. Precision of 3%

vi. Advantages: measuring doses in regions where ion chambers cannot be used

vii. Disadvantages: Cant' get space-dose distribution, Temp and Energy sensitive, can't get immediate readout

When a material is irradiated, some of the electrons in the valence band (ground state) receive enough energy to be raised to the conduction band. The vacancy created in the valence band is called a positive hole. The electron and the hole move independently until they recombine (electron returning to ground state) or until they fall into a trap (metastable state)

- Flouorescence: instantaneous emission of light owing to these transitions
- Phosphorescence (delayed fluorescence): emission of light when electron in the trap requires energy to get out of trap and fall to valence band
- Thermoluminescence: if phosphorescence at room temperature is very slow, but can be speeded up with a moderate amount of heating

b. Silicon Diodes: used for relative electron dosimetry
   i. Advantages: higher sensitivity, instantaneous response, small size, ruggedness
   ii. Limitations: energy dependence for photon beams, directional dependence, thermal effects, RT-induced damage

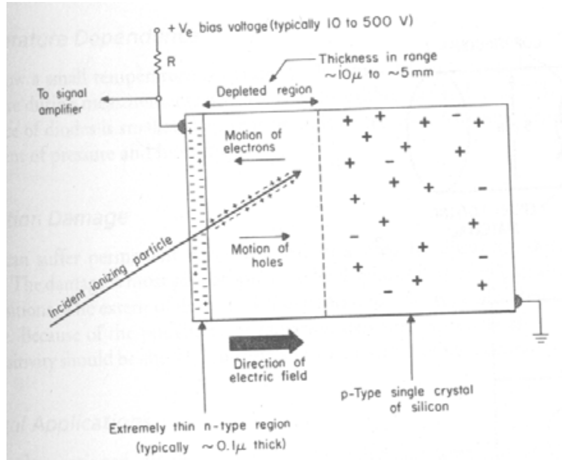


FIG. 8.15. A schematic diagram showing basic design of a silicon p-n junction diode. (From Attix FH. *Introduction to radiological physics and radiation dosimetry.* New York: John Wiley & Sons, 1986, with permission.)

c. Film

i. Radiographic Film: good for relative dosimetry, not for absolute dosimetry

1. Consists of transparent film base coated with crystals of silver bromide. When film irradiated, chemical change takes place in exposed crystals to form a latent image, and when film developed, affected crystals are reduced to small grains of metallic silver (fixed), and appear black on film. Unaffected crystals are removed by the fixing solution.
2. Degree of blackening measured by determining optical density by a densitometer
3. OD = log (I_o/I_t)
   a. Io = amt of light collected w/o film
   b. It = amt of light transmitted through film
4. In dosimetry: quantity of interest is the net optical density. Plot of net optical density vs. dose is called an H-D curve
5. In megavoltage range of photons, film measures with +/- 3% accuracy

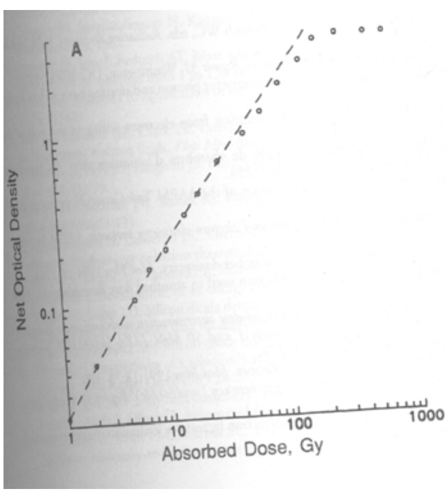


FIG. 8.18. A plot of net optical density as a function of dose for MD-55-2 radiochromic film. (From AAPM. Radiochromic film dosimetry: recommendations of AAPM Radiation Therapy Committee Task Group 55. *Med Phys* 1998;25:2093–2115, with permission.)

ii. Radiochromic film: popular for brachytherapy dosimetry

1. Advantages: tissue equivalence, high spatial resolution, large dynamic range, low energy dependence, insensitive to visible light, no need for chemical processing
2. IR causes colorless film to change to blue

Film badges: With 4 or 5 filtered areas for dose assessment, the appropriate x-ray energy can be determined quite accurately. Film can be analyzed to determine whether or not the exposure was received during an occupational exposure, or, whether or not the film was irradiated in a stationary mode, where the person was not occupationally exposed.

Chapter 9: Dose Distribution and Scatter Analysis

9.1 Phantoms
Dose distribution data is acquired via phantom. Since Compton effect is most relevant for therapeutic radiation, it is necessary that the phantom used have equivalent electron density as that of water. Of commercially available phantoms, Lucite and polystyrene are most frequently used. Solid water phantoms are also used, and have electron density equal to water.

9.3 Percentage Depth Dose
PDD = (D_d/D_{do}) x 100

- D_d = Absorbed dose at any depth, d
- D_{do} = Absorbed dose at a fixed reference depth, do
  - For orthovoltage and lower energy x-rays, do is usually the surface (do = 0)
  - For higher energies, the reference depth is taken at position of peak absorbed dose (do = dmax)

Beam quality:
- PDD increases with beam energy (beyond the depth of dmax)

Dose build-up:

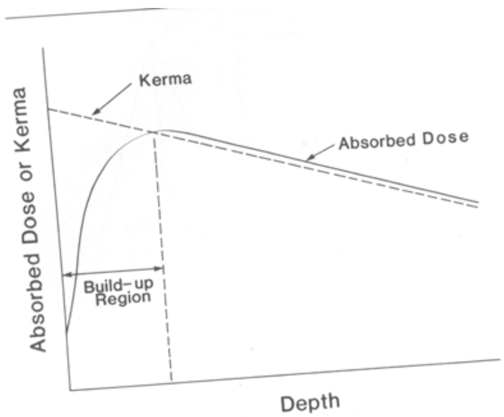


FIG. 9.4. Schematic plot of absorbed dose and kerma as functions of depth.

- with higher energy, more pronounced dose buildup, more skin sparing effect
- Kerma is maximum at surface and decreases with depth b/c of decrease in photon energy fluence
- Absorbed dose first increases with depth and reaches a max at a depth equal to the range of electrons in the medium. Beyond this depth, dose decreases as kerma decreases, resulting in decrease in secondary electron production and net decrease in electron fluence

Effect of Field Size: PDD increases with increasing field size
- As field size increases, contribution of scatter dose to absorbed dose increases
  - Field size dependence is less pronounced for higher energy because there is less random scatter with higher energy

Equivalent square:
- For a given rectangle, the length of the side of the equivalent square is = 4 (area/perimeter)
  - Does not apply to circles or irregularly shaped fields

Equivalent circle:

- Radii = $(4/\sqrt{\pi})$ x (area/perimeter)

Dependence on SSD:
- PDD increases with SSD b/c of the effects of the inverse square law
  - Dose rate decreases slower with distance as the SSD increases

Mayneord F factor: factor used to calculate change in PDD as SDD changes
- Where f = SSD and d = depth

  - F = $(f2 + dm)^2$ x $(f1 + d)^2$

------------   -----------
(f1 + dm)   (f2 + d)

9.4 Tissue-Air Ratio (TAR)
TAR removes SSD dependence. Defined as the ratio of the dose at a given pt in the phantom (D_d) to dose in free space (D_{fs}) at the same pt. Most useful in calculations involving isocentric techniques

$$TAR = Dd/Dfs$$

TAR varies with energy, depth, and field size very much like PDD

Backscatter Factor (BSF) = TAR at depth of dmax

$$BSF = Dm/Dfs$$

Relationship between TAR and PDD:

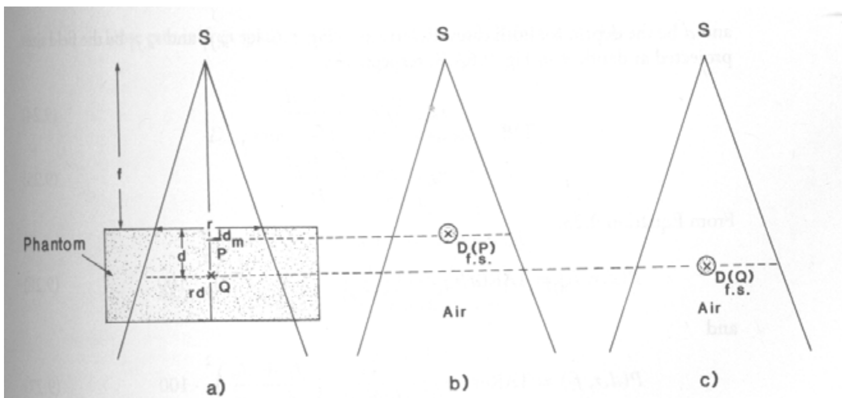


FIG. 9.9. Relationship between TAR and percent depth dose. (See text.)

$$PDD(d,r,f) = TAR(d,rd) \times \frac{1}{BSF(r)} \times \left(\frac{f + dm}{f + d}\right)^2 \times 100$$

How to convert PDD from SSD to another using TAR:

$$\frac{PDD\ (d,r,f2)}{PDD\ (d,r,f1)} = \frac{TAR\ (d,rd,f2)}{TAR\ (d,rd,f1)} \times F$$

To calculate dose in rotation therapy, need to determine the average TAR, and use this value in calculations.

9.5 Scatter-Air Ratio (SAR)
Used to calculate scattered dose in the medium. Defined as ratio of scattered dose at a given pt in the phantom to the dose in free space at the same point. Useful in dosimetry of irregular fields.

- SAR = Dd/Dfs
  - Equal to total dose minus primary dose at a point
  - SAR(d,rd) = TAR(d,rd) – TAR(d,0)

Like TAR, independent of SSD, but depends on beam energy, depth and field size

How to calculate dose in irregular fields---Clarkson's Method

Based on the principle that scattered component of dose (which is depends on field size and shape) can be calculated separately from the primary component (which is independent of field size and shape).

Miscellaneous

- BSF is minimum at energy of 8 MV
- Dmax for various energies
    - Co-60 = 0.5 cm
    - 4 MV = 1 cm
    - 6 MV = 1.5 cm
    - 8 MV = 2 cm
    - 10 MV = 2.5 cm
    - 15 MV = 3.5 cm
    - 25 MV = 4 cm

Chapter 10—A System of Dosimetric Calculations

Problems with dose calculations:

- PDD method: not good with isocentric techniques
- TAR method: may not be accurate for energies higher than Co-60

TMR (Tissue-maximum ratio): used to overcome limitations of TAR

10.1 Dose Calculation Parameters

Effective Primary Dose = dose due to primary photons as well as scatter dose from collimators (excludes scatter dose from phantom)

Collimator Scatter Factor (Sc): ratio of output in air for a given field to that for a reference field

Phantom Scatter Factor (Sp): takes into account the change in scatter originating in the phantom at a reference depth as field size is changed. It is the ratio of the dose rate for a given field at a reference depth to the dose rate at the same depth for the reference field size, with the same collimator opening.

Tissue-Phantom and Tissue-Maximum Ratios:

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

TMR: special case of TPR, and defined as the ratio of the dose at a given pt in phantom to the dose at the same pt at the reference depth of maximum dose.

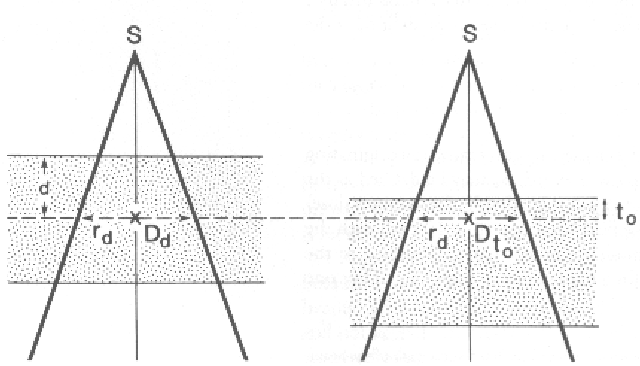


FIG. 10.2. Diagram illustrating the definitions of tissue-phantom ratio (TPR) and tissue-maximum ratio (TMR). TPR(d, r_d) = D_d / D_{t0}, where t_0 is a reference depth. If t_0 is the reference depth of maximum dose, then TMR(d, r_d) = TPR(d, r_d).

$$TMR(d, rd) = \frac{TAR(d, rd)}{BSF(rd)}$$

Scatter-Maximum Ratio (SMR): quantity used to calculate scattered dose in a medium. Defined as ratio of the scattered dose at a given pt in phantom to the effective primary dose at the same point at the reference depth of maximum dose.

10.2 Practical Applications

<u>SSD Technique:</u>

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

- TD = tumor dose
- K = output factor
- d = depth
- r = field size
- r_c = collimator field size
- SSD factor = (SCD/(SSD + to))^2
  - SCD: source to calibration pt distance
  - to = reference depth

<u>Iscocentric Technique:</u> TMR is the quantity of choice for dosimetric calcs for isocentric techniques

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

- ID = isocenter dose
- SAD factor = (SCD/SAD)^2

<u>Asymmetric Fields:</u> To perform MU calcs for asymmetric fields, for either SSD or isocentric techniques, multiply the denominator by the OAR (Off-Axis Ratio)

- OAR: ratio of the primary dose at the off-axis point of interest to the primary dose at the central axis at the same depth for a symmetrically wide open field.

10.3 Other Practical Methods of Calculating Depth Dose Distribution

<u>Irregular Fields:</u> accomplished with Clarkson's method. However, this is time consuming. Can also be calculated using an approximation method. Draw approximate rectangles to define the irradiated areas. This rectangle is called the effective field, while the unblocked field defined by the collimator is called the collimator field. Can then proceed with calculations for the effective field.

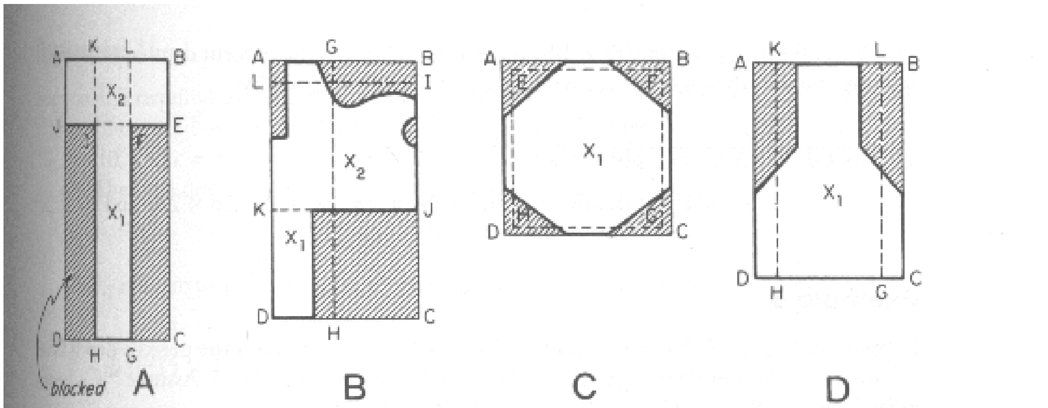


FIG. 10.8. Examples of irregularly shaped fields. Equivalent rectangles for dose at points of interest are shown by dashed lines. Points versus equivalent rectangles are **(A)** 1, *GHKL;* 2, *ABEJ;* **(B)** 1, *AGHD;* 2, *LIJK;* **(C)** 1, *EFGH;* **(D)** 1, *KLGH*. From Levitt SH, Khan FM, Potish RA, eds. *Technological basis of radiation therapy: practical and clinical applications,* 2nd ed. Philadelphia: Lea & Febiger, 1992:73, with permission.

Point Off-Axis: again, can be done with Clarkson's method. However, can be approximated using Day's method

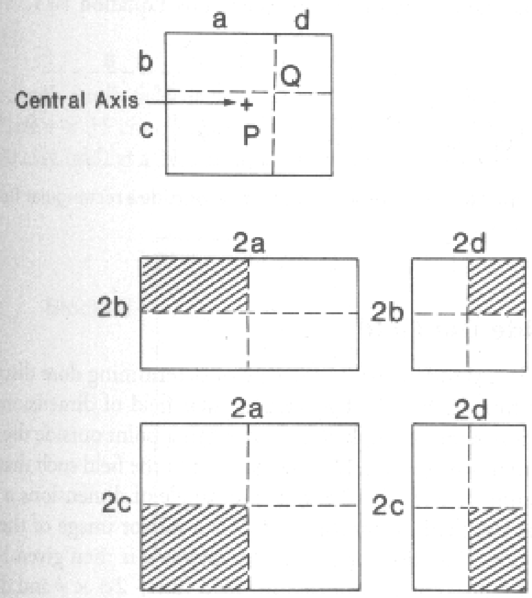


FIG. 10.9. Day's method of calculating dose at any point Q in a rectangular field. (See text.)

To calculate dose at point Q, field can be divided into 4 sections, and the contribution is computed separately.

Dose at depth d along the axis through Q will be given by 1/4 (sum of central axis dose at depth d for fields 2a x 2b, 2a x 2c, 2d x 2b, and 2d x 2c)

If Kq is the off-axis ratio determined in air from the primary beam profile, and if the BSF and central axis PDD for rectangular fields are available, then the dose at depth d along the axis through Q will be given by:

$$\frac{K_q \times 100}{4} \quad \times \quad (\text{sum of BSF} \times \%DD \text{ at depth d for fields } 2a \times 2b,\ 2a \times 2c,\ 2d \times 2b,\ 2d \times 2c)$$

Point Outside the Field: Q is a point outside the field at distance c from field border. Create mirror image to make field symmetric. Dose at pt Q at depth d is given by subtracting the depth dose at Q for field 2c x b from that for field (2a + 2c) x b and dividing by 2.

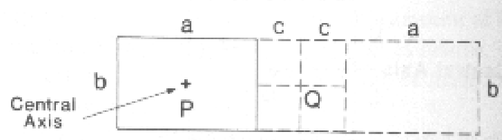


FIG. 10.10. Calculation of depth dose outside a rectangular field. (See text.)

Point Under the Block: Clarkson's method works. Alternatively, can use negative field method.

Dose at pt under block is equal to dose from overall unblocked field – dose expected if entire field were blocked, leaving the shielded volume open. The blocked portion of the field is considered a negative field and its contribution is subtracted from the overall field dose distribution.

Read Indrin's notes to understand this

Chapter 11---Treatment Planning I: Isodose Distributions

11.1 Isodose Chart

- Dose at depth is greatest on central axis, and decreases towards the edges
  - Exception: some x-ray beams display "horns" near the surface in the periphery of the field. These horns are created by a flattening filter, which is used to overcompensate near the surface in order to obtain flat isodose curves at greater depths
- Penumbra region: near edges of the beam. Dose rate decreases rapidly as a function of lateral distance from beam axis
- Near beam edge, falloff caused not only by geometric penumbra, but also by reduced side scatter
  - Physical penumbra: measure of beam sharpness near edges
    - Defined as lateral distance between 2 specified isodose curves at a specified depth
- Field size: defined as lateral distance between the 50% isodose lines at a reference depth

Dose reduction at profile shoulder is due to 2 effects: loss of charged particle equilibrium at the field edge, and beam penumbra (a combination of geometric penumbra and penumbra due to transmission and scattering off the collimator edges---transmission penumbra)

- combined geometric and transmission penumbra = physical penumbra

11.3 Parameters of Isodose Curves

Beam Quality: Depth of a given isodose curve increases with beam quality. Absorbed dose in medium outside primary beam is greater for low-energy beams (due to lateral scatter), causing a greater bulging out of low isodose curves for lower energy beams.

- one potential disadvantage of orthovoltage is increased scatter dose to tissue outside treatment region
- For MV beams, scatter outside the field is minimized as a result of forward scattering

Source size, Source to Surface Distance, and Source to Diaphragm Distance---The Penumbra Effect: these factors affect the shape of the isodose curves by virtue of geometric penumbra (see Chapter 4)

Collimation and Flattening Filter: Function of flattening filter is to make the beam intensity distribution uniform across the field. Filter is thickest in middle, and tapers off toward the edges. Beam flatness is usually specified at a 10-cm depth. To obtain flatness at 10 cm, an area of high dose near the surface may have to be accepted (horns). This is seen more commonly in lower energy beams.

Field Size: A certain isodose curve enclosing the treatment volume should be the guide in choosing a field size rather than the geometric dimensions of the field.

11.4 Wedge Filters

Wedge trays are always at a distance of at least 15 cm from the skin surface, to preserve the skin-sparing effect of the MV beam

Wedge Isodose Angle: the angle through which an isodose curve is tilted at the central ray of a beam at a specified depth (current recommendation is to use depth of 10 cm).

Wedge Transmission Factor: Presence of wedge filter decreases the output of the machine. This must be taken into account in trt calcs. Wedge transmission factor = ratio of doses with and without the wedge.

Wedge Systems:

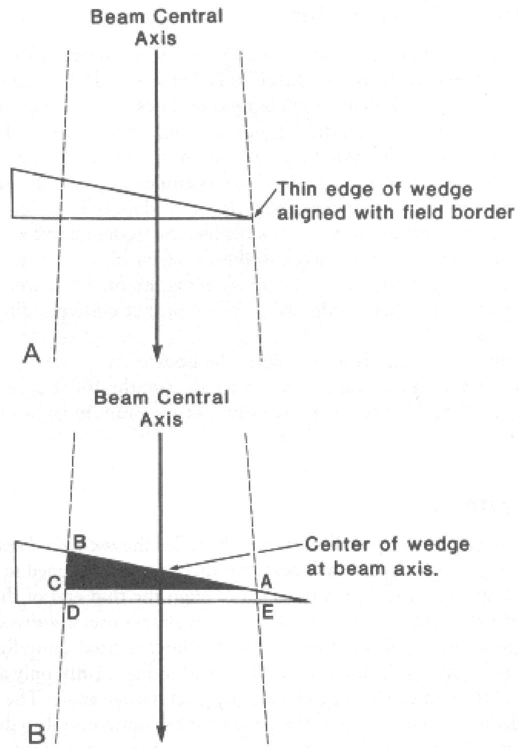


FIG. 11.8. Schematic representation of **(A)** an individualized wedge for a specific field width in which the thin end of the wedge is always aligned with the field border and **(B)** a universal wedge in which the center of the wedge filter is fixed at the beam axis and the field can be opened to any width.

- Individualized wedge system: requires a separate wedge for each beam width (designed to minimize loss of beam output).
    - Thin end of wedge aligned with field border
- Universal wedge: single wedge that works for all beam widths.
    - Filter is fixed centrally in the beam
    - Not as efficient in minimizing loss of output
        - Individualized wedge good for Co-60, where maximizing output is impt
        - Universal wedges are good for linac beams, where output is plentiful

Effect on Beam Quality: Filters can cause beam hardening. For Co-60 beams (monoenergetic), this is not much of an issue. Beam hardening can occur with x-rays. However, the hardening is not significant enough to alter other calculation parameters.

11.5 Combination of Radiation Fields

Parallel Opposed Fields: Uniformity of dose distribution depends on pt thickness, beam energy, and beam flatness. As pt thickness increases or beam energy decreases, the central axis maximum dose near the surface increases relative to the midpoint dose. This effect is called the **tissue lateral effect**.

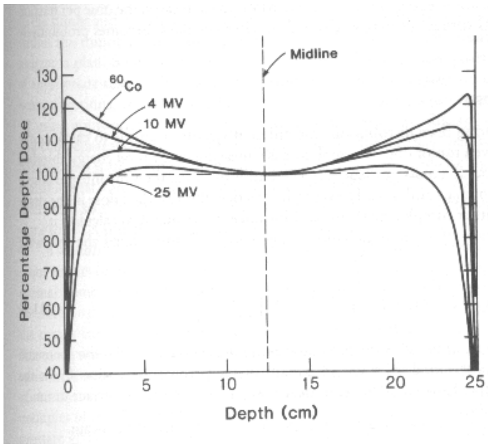


FIG. 11.11. Depth dose curves for parallel opposed field normalized to midpoint value. Patient thickness = 25 cm, field size = 10 × 10 cm, SSD = 100 cm.

Edge Effect (Lateral Tissue Damage): For parallel opposed beams, treating 2 fields/day is better than treating alternating fields daily, even though eventual total dose is same. More biologic damage to normal subcutaneous tissue when it receives alternating high and low doses.

<u>Multiple Fields:</u> can assist in reducing dose to normal tissue surrounding tumor

11.6 Rotational Therapy

11.7 Wedge Field Techniques
Relatively superficial tumors can be irradiated by two wedged beams directed from the same side of the patient.

- Hot spots occur under the thin ends of the wedges of a wedge pair. Magnitude increases with field size and wedge angle.
- Wedge pairs suitable for tumor 0-7 cm deep and when it is necessary to irradiate from one side of the skin surface
- Rapid dose falloff beyond region of overlap

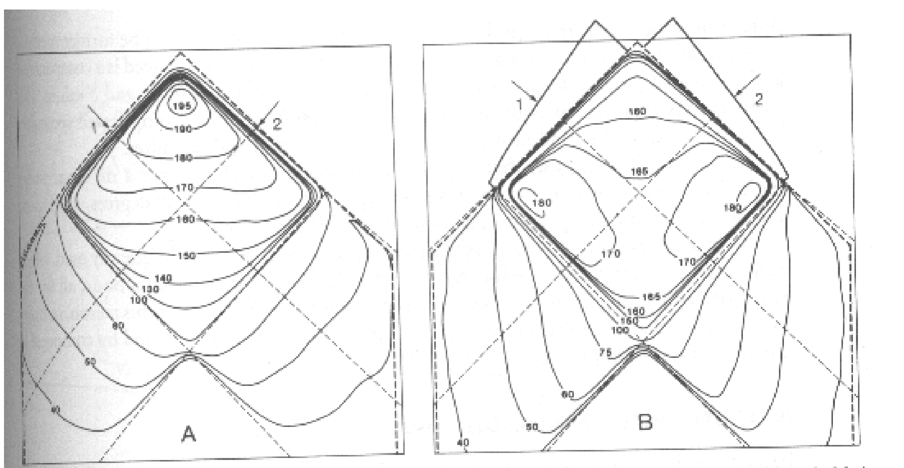


FIG. 11.17. Isodose distribution for two angle beams. **A:** Without wedges. **B:** With wedges; 4 MV, field size = 10 × 10 cm, SSD = 100 cm, wedge angle = 45 degrees.

Wedge angle = 90 – (Hinge angle/2)

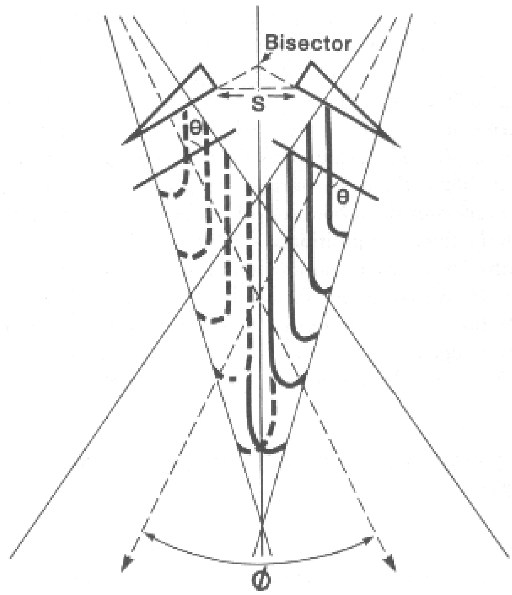


FIG. 11.18. Parameters of the wedge beams, θ is wedge angle, ϕ is hinge angle, and S is separation. Isodose curves for each wedge field are parallel to the bisector.

Uniformity of Dose Distribution: ***Because wedge pairs are normally used for treating small, superficial tumors, high dose region of up to +10% within trt volume is usually acceptable.***

11.8 Tumor Dose Specification for External Photon Beams

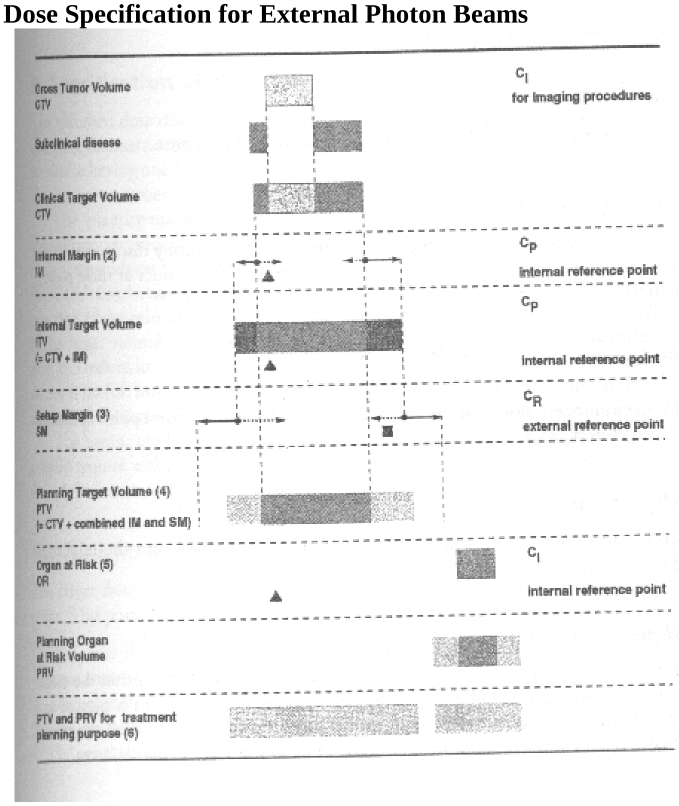


FIG. 11.21. Schematic representation of ICRU volumes and margins. (From ICRU. Prescribing, recording and reporting photon beam therapy [supplement to ICRU Report 50]. ICRU Report 62. Bethesda, Maryland. International Commission on Radiation Units and Measurements, 1999.)

Note: a hot spot is clinically meaningful only if it covers at least an area of 2 cm^2

Chapter 12: Treatment Planning II: Patient Data, Corrections, and Set Up

CT Scan: Insert Figure 12-3

CT Numbers range from –1000 (air) to 1000 (bone). 0 is water.

$$\textbf{H (Hounsfield Number)} = \frac{\mu_{tissue} - \mu_{water}}{\mu_{water}} \times \textbf{1,000}$$

The most basic difference between CT and MRI is that CT is related to electron density and atomic number, while MRI shows proton density distribution.

MRI:

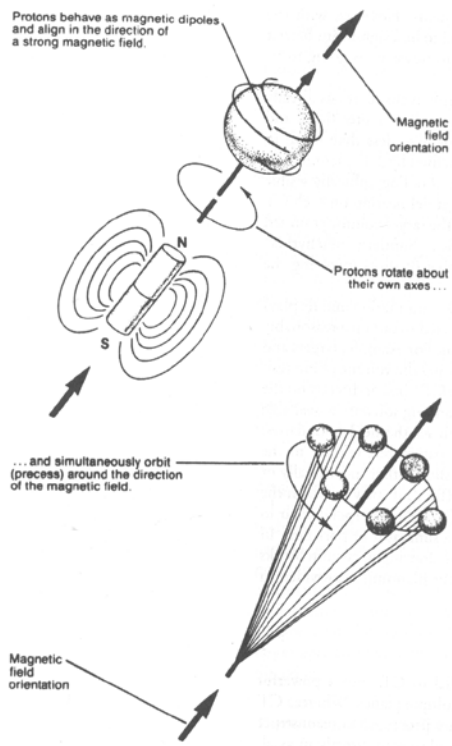


FIG. 12.6. Alignment and precession of protons in a strong magnetic field. (From Halverson RA, Griffiths HJ, Lee BCP, et al. Magnetic resonance imaging and computed tomography in the determination of tumor and treatment volume. In: Levitt SH, Khan FM, Potish RA, eds. *Technological basis of radiation therapy*, 2nd ed. Philadelphia: Lea & Febiger, 1992:38, with permission.)

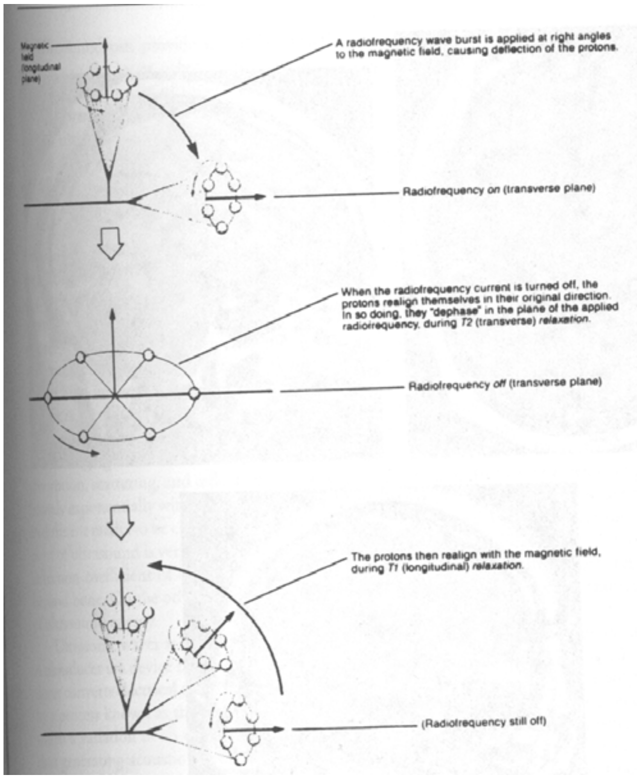


FIG. 12.7. Effects of radiofrequency applied at right angles to the magnetic field. (From Halverson RA, Griffiths HJ, Lee BCP, et al. Magnetic resonance imaging and computed tomography in the determination of tumor and treatment volume. In: Levitt SH, Khan FM, Potish RA, eds. *Technological basis of radiation therapy,* 2nd ed. Philadelphia: Lea & Febiger, 1992:38, with permission.)

Compared to CT: lower spatial resolution, inability to image calcifications.

- Larmor Frequency: frequency of precession.
- MRI based on proton density and proton relaxation characteristics of different tissues.
- To reconstruct the image, use phase encoding and frequency encoding.
- MRI uses spin echo technique in which a 180-degree RF pulse is applied after the initial 90-degree pulse, and resulting signal is received at a time that is = to 2x the interval btwn the two pulses. This is called the Echo Time (TE).
- Time between each 90-degree pulse is called the Repetition Time (TR)
  - T1-weighted image: short TR and TE
  - T2-wt image: long TR and TE
  - Proton (spin) density-weighted image: Long TR and short TE

Ultrasound:

- An ultrasonic wave is a sound wave having a frequency > 20,000 cycles/sec
  - Ultrasound waves of 1-20 MHz are used in diagnostic radiology
  - Piezoelectric effect: process by which an ultrasonic transducer converts electrical energy into ultrasound energy, and vice versa
- Three modes of display
  - A mode (amplitude)
  - B mode (brightness): In RT, cross-sectional info used for treatment planning is derived from B images.
  - M mode (motion): displays the motion of internal structures

12.2 Treatment Simulation
12.3 Treatment Verification

Port Films

- Image quality poor due to megavoltage energy
- For optimum resolution, use single emulsion film with a front lead screen and no rear screen
- Limitations
  - Viewing delay 2/2 processing

        - Poor quality, esp with > 6 MV
        - Impractical to film before every trt

Electronic Portal Imaging Device (EPID)

- Advantage of viewing image instantaneously

12.4 Corrections for Contour Irregularities

- Effective SSD Method
- TAR or TMR Method
- Isodose Shift Method

12.5 Corrections for Tissue Inhomogeneities

- Corrections for Beam Attenuation and Scattering
    - TAR Method
    - Power Law Tissue-air Ratio Method
    - Equivalent TAR Method
    - Isodose Shift Method
    - Typical Correction Factors

- Absorbed Dose within an Inhomogeneity

12.6 Tissue Compensation

Placing bolus directly on skin surface is alright for orthovoltage radiation, but for higher-energy beams results in loss of skin sparing. For these situations, compensating filter should be used, which approximates the effect of bolus as well as preserves skin sparing.

- To preserve skin sparing, compensator is placed $\geq$ 20 cm away from patient's skin

The required thickness of a tissue-equivalent compensator along a ray divided by missing tissue thickness along the same ray = density ratio, or thickness ratio

- This ratio is primarily a function of distance

Compensator cannot be designed to absorbed dose compensations at all depths

- Will overcompensate at shallow depths, and undercompensate at deeper depths
- Average value of 0.7 for the thickness ratio can be used of all RT conditions, provided that d $\geq$ 20 cm

3-D Compensators

- Moire Camera
- Magnetic Digitizer
- CT-based Compensator Systems

Compensating Wedges: A compensating wedge is different from a wedge filter

- Wedge filter: primarily used to tilt the standard isodose curves through a certain wedge angle
    - Wedge filter isodose curves must be available and used to obtain the composite isodose curves before the filter is used in a treatment setup
- Compensating wedge: used just as a compensator
    - Standard isodose charts can be used without modification
    - No wedge transmission factors are required
    - Advantage: can be used for partial field compensation to compensate part of the contour
        - Wedge filter could not do this since it is designed to be placed in the field in a fixed position

12.7 Patient Positioning

- Skin marks should not be relied on for daily positioning. Field boundaries should be defined relative to bony landmarks established during sim
- XYZ Method of Isocenter Set-up: isocenter placed inside the patient, usually at the center of the target volume

- Treatment fields simulated using AP and lat radiographs, and isocenter is established
- Reference anatomic point is chosen, somewhere close to treatment area, to represent a stable anatomic landmark
- Coordinates of the treatment isocenter are recorded

Chapter 13---Treatment Planning III---Field Shaping, Skin Dose, and Field Separation

13.1 Field Blocks

Block Thickness: Primary beam transmission of 5% through a block is considered acceptable in most situations (between 4.5-5 HVL of lead). Shielding for superficial and orthovoltage beams can be done with thin sheets of lead placed or molded onto skin surface. For megavoltage beams, thickness of shielding increases substantially, and blocks are placed above the patient, supported in the beam on a shadow tray.

Block Divergence: Ideally, blocks should be shaped or tapered so that their sides follow the geometric divergence of the beam. This minimizes block transmission penumbra (partial transmission of the beam at the edges of the block). However, this offers little advantage for beams with large geometric penumbra, where straight-cut blocks are acceptable. Divergent blocks are best used for beams with small focal spots.

13.2 Field Shaping

Custom Blocking: ***Cerrobend is commonly used to make custom blocks. It has density that is 83% of lead. Advantage is that its melting pt is much lower than lead, so it is easier to cast into shapes. For megavoltage beams, commonly use 7.5 cm blocks (equivalent to 6 cm lead).***

* Thickness of cerrobend needed = 0.6 mm per MeV (or 1.2x the thickness of lead)

Independent Jaws: Rectangular blocking can be accomplished with independent jaws or independently movable collimators. Convenient when matching fields or beam splitting (where beam can be blocked at central axis to remove divergence). Asymmetric collimation can change the physical penumbra, with tilt of isodose curves towards blocked edge (blocking results in elimination of photon and electron scatter from blocked portion of the field, reducing the dose near the edge).

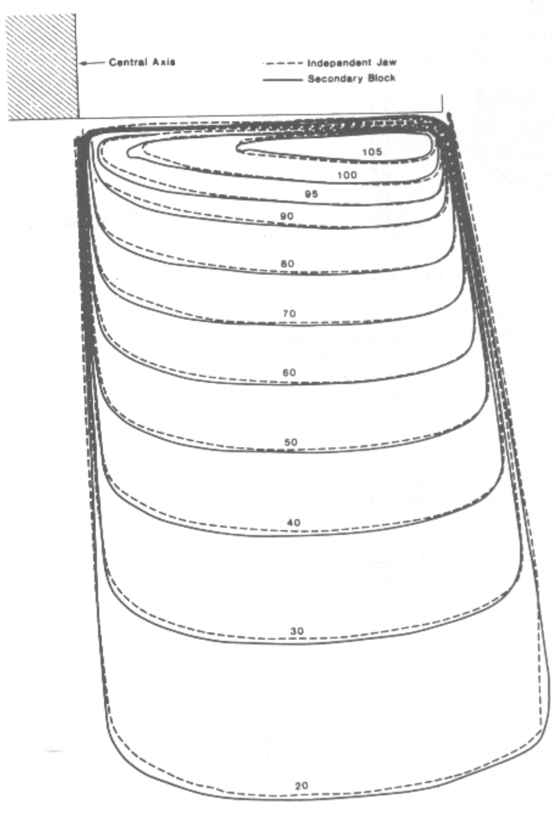


FIG. 13.3. Comparison of isodose distrib with half the beam blocked by an indepe jaw versus a block on a tray. Notice close ment as well as the tilt of the isodose cur ward the blocked edge.

MLC's: Leaves are made of tungsten alloy. Primary x-ray transmission through leaves of < 2%, and interleaf transmission is < 3%. Ideally suited for trts requiring multiple fields (3D-CRT and IMRT).

13.3 Skin Dose

Electron Contamination of Photon Beams: Surface dose is the result of electron contamination of the incident beam as well as the backscattered radiation (electrons and photons) from the medium. These electrons arise from photon interactions in air, in the collimator, and in any other scattering material in the path of the beam.

Measurement of Dose Distribution in the Build-up Region: Extrapolation chambers are the instruments of choice (but few institutions have them). Fixed-separation plane-parallel chambers are used instead commonly. Thin layers of TLD material can also be used for measuring dose distribution in the build-up region. In vivo measurements of surface dose can also be made by placing TLD chips directly on skin surface.

Effect of Absorber-skin Distance: the electron contamination with no absorber placed in the beam is mainly caused by the secondary electron emission from the collimator. When an absorber of thickness greater than the range of secondary electrons is introduced, the collimator electrons are almost completely absorbed but the absorber itself becomes the principal source of electron contamination. By increasing the distance between the tray and the surface, the electron fluence incident on the skin is reduced because of divergence and absorption and scattering of electrons in air. Thus, skin sparing enhanced by placing the shadow tray farther away from skin (15-20 cm).

Beam spoiler: made from a low atomic number material. When placed at an appropriate distance from the surface, can be used to increase the dose superficially. Often used for TBI.

Effect of Field Size: as field size increases, skin sparing decreases. Due to increased electron emission from the collimator and air. When using large fields with a tray-to-skin distance of 15-20 cm, it is necessary to use electron filters to maintain the skin-sparing effect.

Electron Filters: Medium atomic number material (30-80) can be used as electron filters by reducing the secondary eletctron scatter in the forward direction. Optimally happens at Z=50 (tin).

Skin Sparing at Oblique Incidence: Skin dose increases with increasing beam obliquity, and dmax decreases.

13.4 Separation of Adjacent Fields

Methods of Field Separation:

1) Geometric: In the case of 2 adjacent beams, matched at depth d, to calculate the skin gap:

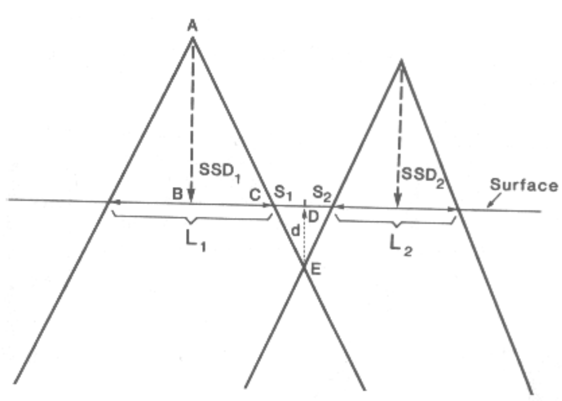


FIG. 13.13. Geometry of two adjacent beams, separated by a distance $S_1 + S_2$ on the surface and junctioning at depth d.

Total separation on skin surface = S1 + S2 = (1/2) L1 * d/SSD1 + (1/2) L2 * d/SSD2

Certain arrangements cause regions of "three-field overlap"

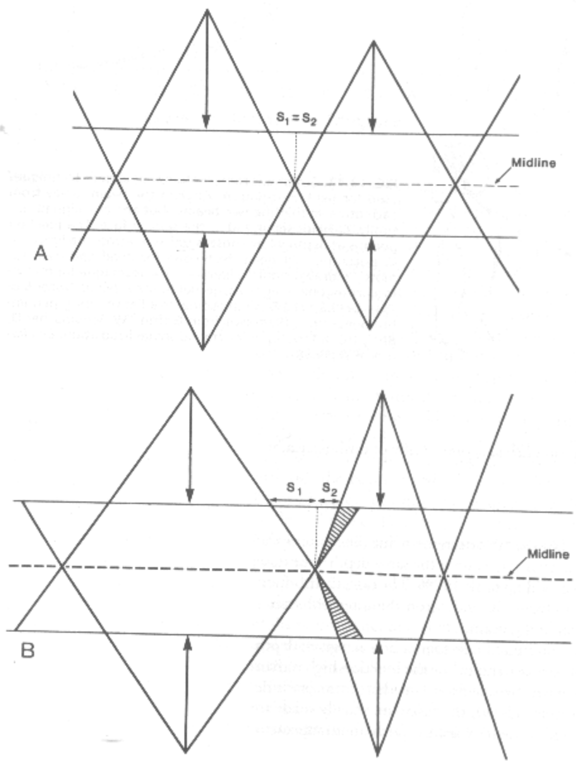


FIG. 13.14. Two pairs of parallel opposed fields. Adjacent fields are separated on the surface so that they all join at a point on the midline. **A:** Ideal geometry in which there is no three-field overlap. **B:** Arrangement in which there are two regions (shaded) of three-field overlap.

The length of the three-field overlap on the surface is given by S1-S2

- The length of the overlap can be made 0 if L1/L2 = SSD1/SSD2
    - Thus if field lengths are different, can adjust SSD's to eliminate overlap.

2) Dosimetric: Orthogonal Field Junctions (Example: CSI)

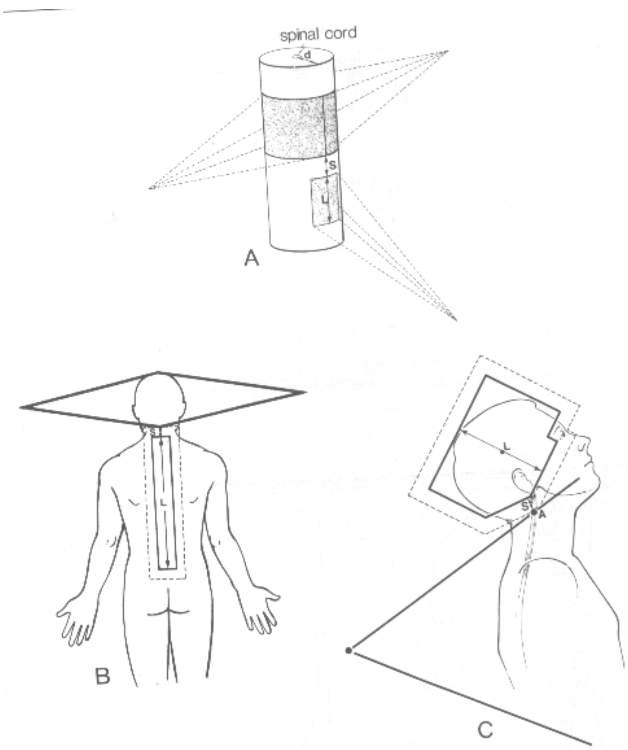


FIG. 13.16. A: A general diagram showing the separation of orthogonal fields. **B:** An example of orthogonal fields used for craniospinal irradiation. **C:** A lateral view of B, illustrating the geometry of orthogonal field separation.

The junction between the cranial and spinal fields can be accomplished in several ways:

Technique 1:

Separation (S) = (1/2) L * d/SSD

- L is the length of the spinal field
- d = depth of spine from the posterior surface
- SSD = SSD for the spinal field

Angle of collimator rotation = arc tan (1/2 * Length of posterior spinal field * 1/SSD of spinal field)

Angle of couch rotation = arc tan (1/2 * Length of lateral cranial field * 1/SAD of cranial field)

- the couch is rotated toward the side the cranial field enters the head

Technique 2:

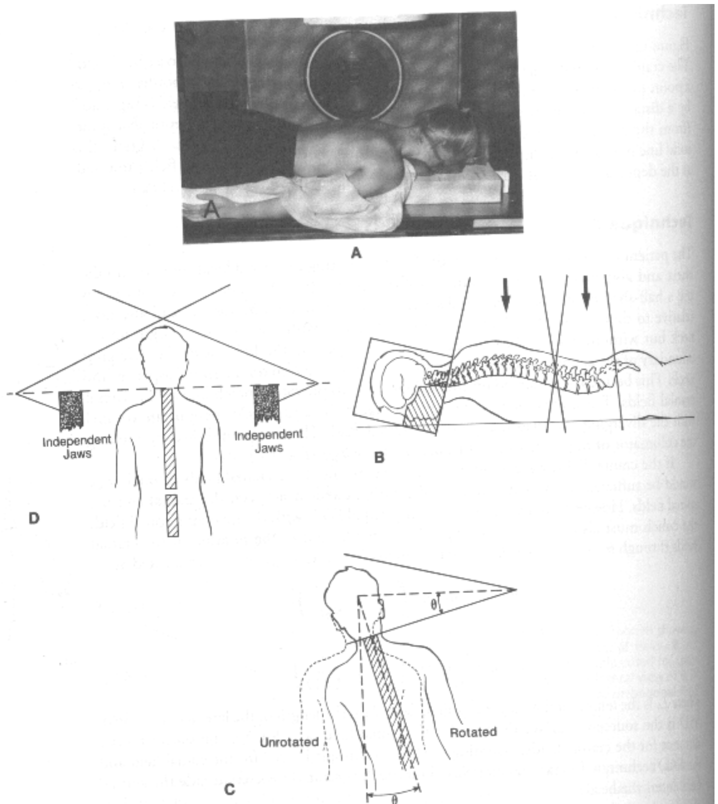


FIG. 13.17. Craniospinal irradiation technique. A: Patient set-up showing Styrofoam blocks and Alpha Cradle mold to provide stable position for abdomen, chest, and head. B: Lateral view of fields showing cranial field rotated to align with the diverging border of the spinal field. C: Couch rotated to provide match between the spinal field and the diverging border of the cranial field. D: Elimination of cranial field divergence by using an independent jaw as a beam splitter. This provides an alternative to couch rotation in C.

Alternative approach to rotating the couch is to eliminate cranial field divergence by using a half-beam block or an independent jaw to split the fields at the craniospinal junction line. The beam splitter is positioned at the central axis or close to it, to eliminate divergence of the rays at the junction line. The collimator is still rotated as described above, but the couch does not need to be rotated.

- Advantages:
  - Orthogonal field matching is achieved with no overlaps between the cranial and spinal fields at any depth
  - Independent jaw can be used to move the junction line by a centimeter each week during the treatment course to smear out the junctional dose distribution

Guidelines for Field Matching

1) Site of field matching should be chosen to be over an area that does not contain tumor or a critically sensitive organ.
2) If the tumor is superficial at the junction site, fields should not be separated because of risk of creating a cold spot in the tumor. However, if the diverging fields abut on the skin surface, they will overlap at depth. This may be clinically acceptable, provided that excessive dose delivered to underlying tissues does not exceed their tolerance. In the case of

a superficial tumor with a critical organ located at depth, one may abut fields at the surface but eliminate beam divergence using a beam splitter or by tilting the beams.

3) For deep-seated tumors, the fields may be separated on the skin surface so that the junction point lies at the midline.
4) The line of field matching must be drawn at each treatment session on the basis of the first field treated. It is not necessary anatomically to reproduce this line every day because variation in its location will smear the junction point, which is desirable. For the same reason some advocate moving the junction site two or three times during a treatment course

Chapter 14---Electron Beam Therapy

Electron Scattering: When electrons pass through a medium, electrons suffer multiple scattering due to Coulomb force interactions between electrons and the nuclei of atoms.

- High Z materials produce more scatter.
  - Scattering foils use high Z material
- Increasing kinetic energy of electrons = less scatter

Energy Specification and Measurement: In clinical practice, electron beams characterized by energy at the body surface

- ***Most Probable Energy*** $(E_p)_0$**:** defined at the phantom surface
  - $(E_p)_0 = C_1 + C_2R_p + C_3R_p^2$
    - R_p = practical range of electron
      - Depth of the point where the tangent to the descending linear portion of the curve intersects the extrapolated background

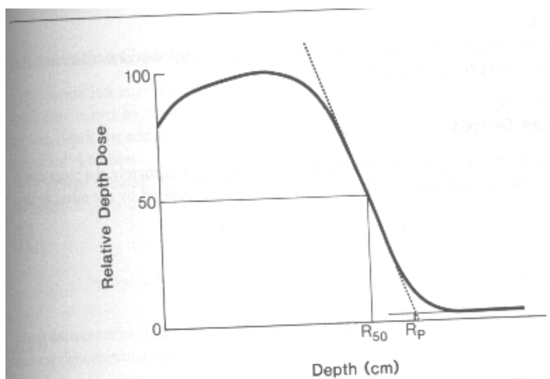


FIG. 14.3. Depth-dose curve illustrating the definition of R_p and R_{50}.

- ***Mean Energy*** $(\overline{E}_0)$**:**
  - $= C_4 \times R_{50}$
    - C4 = 2.33 MeV/cm

- ***Energy at Depth:*** The most probable energy and the mean energy both decrease linearly with depth
  - E_d (MeV) $= E_o (1 - d/R_p)$
    - d = depth in phantom

- ***Ranges:***
  - Range of electron beam = (1/2) x E
  - R90 = (1/3.2) x E
  - R80 = (1/2.8) x E

Determination of Absorbed Dose: Depth-dose and isodose distributions can be determined by ion chambers, diodes, or film (film restricted to relative dosimetry). Water phantoms are standard for electron beam dosimetry, but since not always feasible to use water, polystyrene and electron solid water come closest to being water equivalent.

Characteristics of Clinical Electron Beams:

Central Axis Depth-Dose Curves:

- High energy electrons lose energy at 2 MeV/cm of water or soft tissue. Beyond the maximum range of electrons, the dose is contributed only by the x-ray contamination of the beam.
- The most useful treatment depth is given by the depth of the 90% depth dose
- Surface dose increases with energy
    - At lower energies, electrons are scattered through larger angles
    - Ratio of surface dose:maximum dose is less for lower-energy electrons

Isodose curves:

- For low energy beams: all isodose curves bulge out (b/c low energy beams undergo greater scattering angles)
- For high energy beams: low isodose levels bulge out, higher isodose levels show lateral constriction

Field Flatness and Symmetry:

- scattering foils (usually made of lead) and beam defining collimators are used to give acceptable field flatness and symmetry.
    - Scattering foils widen the beam and give a uniform dose distribution across the treatment field
    - Collimators:
        - Primary collimation close to source that defines max field size
        - Secondary collimation close to patient to define trt field

Field Size Dependence:

- Output and central-axis depth dose are field size dependent, if CPE is not maintained
- CPE maintained as long as total field size = range of secondary electron
    - If field size is < range of electrons, the CPE is lost, and PDD decreases
        - Depth of dmax shifts toward the surface for smaller fields

Electron Source:

- Virtual source: intersection point of the backprojections along the most probable directions of electron motion at the patient surface.

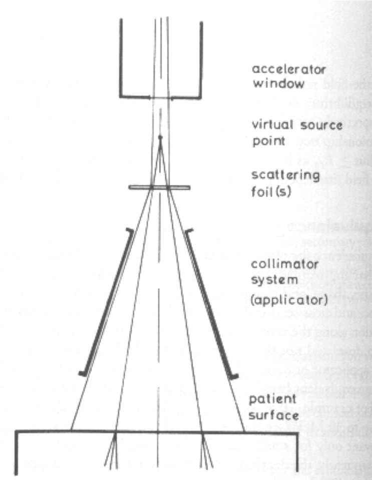


FIG. 14.18. Definition of virtual point source of an electron beam: the intersection point of the backprojections along the most probable directions of motion of electrons at the patient surface. (From Schroeder-Babo P. Determination of the virtual electron source of a betatron. *Acta Radiol* 1983;364[suppl]:7, with permission.)

- Virtual SSD: does not give accurate inverse square law correction for output at extended SSDs under all clinical conditions
    - Virtual SSD gives correct inverse square law factors only for large field sizes
    - For small field sizes, inverse square law correction underestimates the change in output with virtual SSD
- Effective SSD: alternative method of correcting dose output for the air gap between the electron collimator and patient
    - Gives the correct inverse square law relationship for the change in output with distance

X-ray contamination:

- Accounts for the tail at the end of the depth-dose curve
    - Due to bremsstrahlung interactions of electrons with the collimation system and the body tissues

- % of x-ray contamination increases with increasing electron energy

Treatment Planning

Choice of Energy and Field Size:

- In general, set beam energy so that target lies entirely within the 90% IDL
- With breast, energy often chosen so that IDL at chest wall-lung interface is 80% (to spare lung)
- Choice of field size: strictly based on isodose coverage of the target volume

Beam obliquity:

- With increased obliquity:
    - Dmax shifts towards surface
    - Deceased depth of penetration
    - Surface dose increases

Tissue inhomogeneities:

- Dose distribution beyond the inhomogeneity can be corrected by using the coefficient of equivalent thickness (CET) method
- The CET for a material is given by its electron density relative to that of water
- Dose at point beyond the inhomogeneity is determined by calculating the effective depth (d_{eff})
    - $d_{eff} = d - z(1 - CET)$
        - d = actual depth of point P from the surface
        - z = thickness of inhomogeneity

Use of bolus and absorbers:

- Bolus is used with electrons to:
    - Flatten out an irregular surface
    - Reduce the penetration of electrons in parts of the field
    - Increase the surface dose

Matching adjacent fields:
When an electron field is abutted at the surface with a photon field, hot spot develops on the side of the photon field, and cold spot develops on the side of the electron field. Method and amount of overlap depends on the clinical scenario.

Field Shaping
Lead cutouts often used to give shape to treatment area and shield surrounding normal tissue

- Cutouts placed on skin or at end of treatment cone
- For e-beams < 10 MeV, less than 5 mm lead will provide adequate shielding ($\leq$ 5% transmission)
- The minimum thickness of lead required for blocking in mm is given by the electron energy (in MeV) incidence on lead divided by 2
    - Required thickness of Cerrobend is 20% greater than that of pure lead

Effect of blocking on dose rate:

- If a field produced by a lead cutout is smaller than the minimum size required for maximum lateral dose buildup, the dose in the open portion is reduced
    - For an irregularly shaped field, field edges should be farther than Rp/2 for lateral scatter equilibrium to be achieved

Internal shielding:

- Used in situations such as lip, buccal mucosa, and eyelid lesions
- Electron backscatter from lead can enhance the dose at the tissue-lead interface
    - Higher value for lower-energy beams
    - Higher Z material = more backscatter
- To dissipate the effect of electron backscatter, a low Z absorber may be placed between the lead shield and preceding tissue surface

Electron Arc Therapy
Gives good dose distribution for treating superficial tumors along curved surfaces.

Beam energy:

- central axis dose distribution is altered due to field motion
  - Beam appears to penetrate somewhat farther than for a stationary beam
  - Surface dose is reduced and bremsstrahlung dose at isocenter is increased (this is called the velocity effect)

Scanning Field Width:

- Geometric field width of 4-8 cm at the isocenter is recommended for most clinical situations

Location of isocenter:

- Should be placed at a point that is equidistant from the surface contour for all beam angles
- Depth of isocenter should be greater than maximum range of electrons, so that there is no accumulation of electron dose at the isocenter

Field shaping:

- lead strips or cutouts should be used to define the arc limits as the field limits in the length direction, to sharpen the distribution

Total Skin Irradiation

Electrons in the energy range of 2-9 MeV useful for treating superficial lesions covering large areas of the body

- Rapid falloff in dose beyond a shallow depth and minimal x-ray background (1% or less)
- Superficial lesions extending to 1 cm depth can be treated effectively without exceeding bone marrow tolerance

Translational technique:

- Patient lies on a motor-driven couch and is moved relative to a downward-directed beam
- Alternatively, patient may be stationary and the radiation source translated horizontally

Large field technique:

- Standing patient is treated with a combination of broad beams produced by electron scattering and large SSDs (2-6 meters)
- X-ray contamination is a limiting factor in total skin irradiation
  - Low energy electron beams are widened by scattering in air
  - If electron beams is scattered by air alone before incidence on patient, this reduces x-ray contamination
  - If a scattering foil is used, x-ray contamination would increase
- Although a dose uniformity of +/- 10% can be achieved over most of the body surface, areas such as inner thighs and axillae, which are obstructed by adjacent body structures, require supplementary irradiation

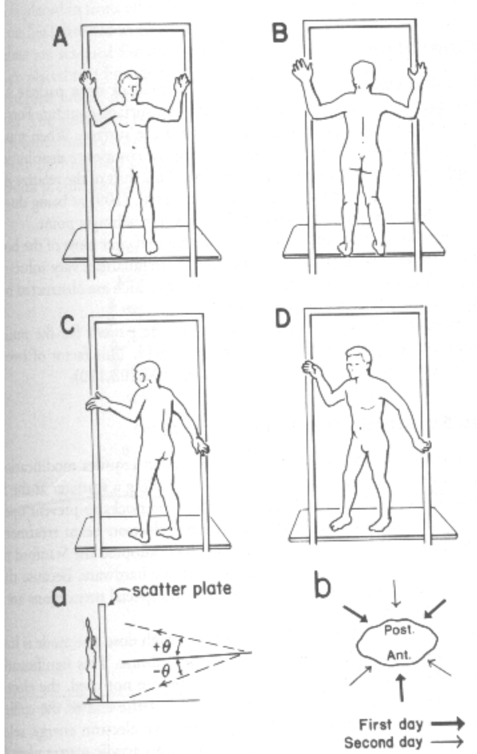


FIG. 14.45. Patient positions for the six-field Stanford technique. Patient is treated by two beams at each position, one beam directed 15 degrees below horizontal and the other 15 degrees above horizontal. (From Page V, Gardner A, Karzmark CJ. Patient dosimetry in the treatment of large superficial lesions. *Radiology* 1970;94:635, with permission.)

In-vivo dosimetry: TLD's are most commonly used

Chapter 15---Brachytherapy

Properties of Commonly Used Radionuclides

Radionuclide	E (MeV)	$T_{1/2}$	Forms	Use	HVL (in mm Pb)
^{226}Ra	0.83 (avg)	1600 y	tubes, needles	IC, IS	8.0
^{222}Rn	0.83 (avg)	3.82 d	seeds	IS	8.0
^{60}Co	1.17, 1.33	5.27 y	tubes, needles	IC, IS	11.0
^{137}Cs	0.662	30.1 y	tubes, needles	IC, IS	5.5
^{192}Ir	0.38 (avg)	73.83 d	seeds in ribbon, wire	IC, IS	2.5
^{198}Au	0.412	2.7 d	seeds	IS	2.5
^{125}I	0.028 (avg)	59.4 d	seeds	IS, IV	0.025
^{131}I	0.61 (β)	8.02 d	liquid	Thyroid	2.3
^{103}Pd	0.021 (avg)	17.0 d	seeds	IS	0.008
^{32}P	0.7 (β)	14.26 d	colloid, wire	Abdominal, cranial, IV	---
^{90}Sr	2.27 (β)	28.79 d	foil, beads	Pterygium, IV	---

IC- intracavitary, IS-interstitial, IV-intravascular

Iodine-125: decays exclusively by electron capture, and then decays to ground state by emission of gamma rays. Characteristic x-rays are produced due to electron capture and internal conversion. Dose distribution around iodine seeds is highly anisotropic. This can produce cold spots, but problem addressed by creating random seed distributions. Most treatment planning systems do not take into account anisotropy around individual sources.

Palladium-103: has a shorter half life than Iodine-125. May provide biologic advantage in permanent implants because the dose is delivered at a much faster rate. Also decays by electron capture with emission of characteristic x-rays and Auger electrons. Like I-125, anisotropy seen in the dose distribution around the source.

Historical review: role of Radium

Radium was most commonly used isotope, but no longer clinically used (High HVL, long $T_{1/2}$, production of by-products).

Source specification for Radium:
1) Active length: length of radioactive material
2) Physical length: length of encapsulated source
3) Activity/Source strength: mgRa (1 mgRa = 1 mCi)
4) Filtration: thickness of encapsulation (mm Pt)

Exposure Rate Constant (Γ) : activity of a nuclide is related to the exposure rate by the exposure rate constant

Γ = exposure rate (R/hr) at a pt 1 cm from a 1 mCi point source
(Γ_{Ra} filtered by 0.5 mm Pt = 8.25 Rcm2/mg hr)

Specification of Sources: strength of a source can be specified in several ways:

1) Activity: source strength may be specified in terms of mCi

$$A(t) = Ao\, e^{-\lambda t}$$

Point Source Approximation of exposure:

$$X = \frac{A \times \Gamma}{d^2}$$

- A = activity
- Γ = exposure rate
- d = distance

2) Exposure Rate at a Specified Distance: NCRP recommends that strength can be specified by exposure rate in air at a specified distance. Units of R/hr, and was until recently the way source strength was specified, now replaced by air kerma strength.
3) Equivalent Mass of Radium: a mgRaEq of source X provides the same exposure rate as a Ra source of specified mass

$$A_{Ra}\,(\text{mgRaEq}) = \frac{A_x(\text{mCi}) \times \Gamma_x}{\Gamma_{Ra}}$$

- if source strength is specified in mgRaEq, use Γ_{Ra} to calculate exposure

4) Apparent Activity: Content activity, activity of an unfiltered point source that has the same exposure rate at 1m in free space as the source. Units = mCi
5) Air Kerma Strength (Sk): 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. ***This is the current recommended quantity for specification of brachytherapy sources.***

$$S_k = K_l \times l^2$$

- K_l is the air kerma rate at a distance l
- Recommended units: $\mu Gy m^2 hr^{-1}$ = 1 U

KERMA consists of:
1) Collisional component: energy dissipated locally due to ionization and excitation in or near electron track
2) Radiative component: energy dissipated in the form of photons (Bremsstrahlung)
3) K = Kc + Kr

Collision kerma in air is the ionization = exposure

$$K_{air} = X \times (W/e)$$

- W/e = average energy required to produce an ion pair in dry air (33.97 J/C)

Therefore,

$$S_k = K_l \times l^2 = X \times (W/e) \times l^2$$

Exposure Rate Calibration: the NIST has established exposure rate calibration standards for brachytherapy sources. Calibration of clinical sources should be directly traceable to NIST or one of the AAPM-ADCLs. Sources should be calibrated by direct comparison with a NIST- or ADCL-calibrated source of the same kind.

1) Open air measurements: used to calibrate sources. Large source to ion chamber distance. Chamber volume should be $\geq$ 100 ml. Time consuming, not suitable for routine calibration checks.
2) Well-type ion chambers: used for routine calibration of brachy sources.

Anisotropy

Point sources: dose at the same distance from the source at different angles will all be the same

Line sources: doses at the same distance from the source will be different due to:
- self absorption within the source
- differences in photon absorption and scatter within the encapsulation

Effect of anisotropy is less obvious at large distances from the source (> 3x length of source)
- At these distances, source can be approximated as a point

Systems of Implant Dosimetry

TABLE 15.8. RULES OF INTERSTITIAL IMPLANT SYSTEMS

Characteristic	Paterson-Parker	Quimby	Paris	Computer[a]
Linear strength	Variable (full intensity, 0.66 mg Ra/cm; half intensity, 0.33 mg Ra/cm)	Constant (full intensity, 1 mg Ra/cm; half intensity, 0.5 mg Ra/cm)	Constant (0.6–1.8 mg Ra eq/cm)	Constant (0.2–0.4 mg Ra eq/cm)
Source distribution	Planar implants: Area <25 cm², 2/3 Ra in periphery; area 25 to 100 cm², 1/2 Ra in periphery. Area >100 cm², 1/3 Ra in periphery Volume implants Cylinder: belt, four parts; core, two parts; each end, one part. Sphere: shell, six parts; core, two parts. Cube: each side, one part; core, two parts.	Uniform Uniform distribution of sources throughout the volume	Uniform Line sources arranged in parallel planes	Uniform Line sources arranged in parallel planes or cylindric volumes
Line source spacing	Constant approximately 1 cm apart from each other or from crossing ends	Same as Paterson-Parker	Constant, but selected according to implant dimensions-larger spacing used in large volumes; 8 mm minimum to 15 mm maximum separation	Constant, 1–1.5 cm, depending on size of implant (larger spacing for larger size implants)
Crossing needles	Crossing needles required to enhance dose at implant ends	Same as Paterson-Parker	Crossing needles not used; active length 30% to 40% longer than target length	Crossing needles not required; active length of sources 30% to 40% longer than target length

[a]The computer system used at the University of Minnesota Hospital.
From Khan FM. Brachytherapy: rules of implantation and dose specification. In: Levitt SH, Khan FM, Potish RA, eds. *Technological basis of radiation therapy.* Philadelphia: Lea & Febiger, 1992:113, with permission.

1) The Patterson-Parker System

- Max dose homogeneity ($\pm$ 10%) inside implanted volume (except localized hot spots around the source)
- Stated dose is 10% above the minimum peripheral dose

2) Quimby System

- non-uniform dose distribution
    - Higher in center than in periphery
- Stated dose is the max in the plane for planar implants, and the min dose within the volume for volume implants

3) Memorial System

- extension of Quimby system
- Sources of uniform strength, spaced 1 cm apart

4) Paris System

- Developed for use with continuous wires of Ir
- Reference dose rate is specified as 85% of basal dose
    - Basal dose rate: average dose rate at pts intermediate between the sources (avg of minimum dose between sources)

5) Computer System

- Sources are spaced uniformly, with large spacing for larger size implants, covering the entire volume
- Non-uniform dose distribution: hotter in middle
- Dose prescribed to isodose surface that just surrounds the target/implant
- Follows this criterion: better to implant a larger volume than to select a lower value isodose curve to increase the coverage

Computer Dosimetry

Allows preplanning with implants, as well as seeing complete isodose distribution, corresponding to the final source distribution. Rapid turnaround time with computer dosimetry allows for modifications to be easily made.

Dose calculation requires spatial coordinates for each radioactive source. 3-D reconstruction of source geometry is accomplished by using a set of 2 radiographs, exposed with either orthogonal or "stereo-shift" geometry.

1) Orthogonal Imaging Method: orthogonal films obtained, and coordinates for each of the sources are assigned
2) Stereo-shift Method: two radiographs of the same view are taken, but the patient or x-ray tube is shifted a certain distance between the two exposures.

- Accuracy of orthogonal method is better
- Stereo-shift method better for cases in which sources cannot be easily identified by orthogonal films

Dose computations:

Dose Rate:
- Dose Rate in Air (Point Source)
    - = X (R/hr) * f
        - f = f factor = roentgen to rad conversion factor
            - = 0.876 cGy/R

- Dose Rate in Air (Line Source): uses Sievert Integral
    - Sievert Integral determines dose rate for a line source, where source is divided into infinitely small sources
        - Inverse square and filtration corrections applied to each source element
            - Limitations of Sievert Integral:
                - Underestimates dose near or along source axis
                - Does not acct for attenuation and scatter in tissue
                    - Scatter compensates for attenuation close to the source, but attenuation dominates farther from the source
                - Does not acct for self absorption within source
                - Breaks down at extreme oblique directions

- Dose Rate in Medium (Point Source)
  - $D_{isotropic} = X * f_{med} * T(r)$
  - $D_{anisotropic} = X * f_{med} * T(r) * \varphi_{an}$

    - T(r) = Tissue Attenuation Factor
    - φ_{an} = anisotropy constant
    - f factor dependent on the medium and energy of photons

- Dose Rate in Medium (Line Source)
  - AAPM-TG 43 equation is what is currently used to calculate dose from a line source, which allows several physical factors to be considered separately such as geometric falloff of photon fluence with distance, anisotropy, and radial dependence of photon absorption and scatter in the medium

Dose:

- Temporary Implants (No Decay)
  - Dose = Dose rate * time

- Permanent Implants
  - Dose = Initial Dose Rate x T_{avg}
    - $T_{avg} = 1.44 * T_{1/2}$

- Temporary Implants (with Source Decay)
  - Dose = Initial Dose Rate * $T_{avg} (1-e^{-t/Tavg})$

Implantation Techniques

Sources can be applied in the following ways:

1) Surface Molds:
2) Interstitial:
3) Intracavitary:

Intracavitary Therapy for Uterine Cervix

Dose Specification: 3 systems used

1) Milligram-Hours: product of total source strength * implant duration. Lacks info on source arrangement, position, tumor size, pt anatomy. Large uncertainties.
2) Manchester System: characterized by 4 pts: A, B, bladder, and rectum. Pt A is the point to which dose is prescribed
3) ICRU System: Dose prescribed to the isodose surface just surrounding the target volume rather than to a point. Implant duration should be based on minimum target dose rate.

The following are the ICRU reference points:

- Point A = 2 cm superior to external cervical os and 2 cm lateral to cervical canal. Represents point where uterine vessels cross the ureter. Problems with using pt A:
  - Pt A relates only to position of sources, not specific anatomy
  - Dose at pt A sensitive to position of ovoids relative to tandem
  - Depending on cervical size, pt A may lie inside or outside the tumor, risking overdosing of small tumors, and underdosing of large tumors.
- Point B = 3 cm lateral to pt A. Represents position of pelvic LN
- Bladder point: localized with Foley with contrast-filled balloon
  - AP film: pt located in center of balloon
  - Lateral film: pt at the center of the posterior surface of the balloon
- Rectal point:
  - AP film: midpt of ovoids
  - Lat film: line drawn from middle of ovoids, 5 mm behind posterior vaginal wall (vaginal wall visualized using radioopaque gauze for packing)

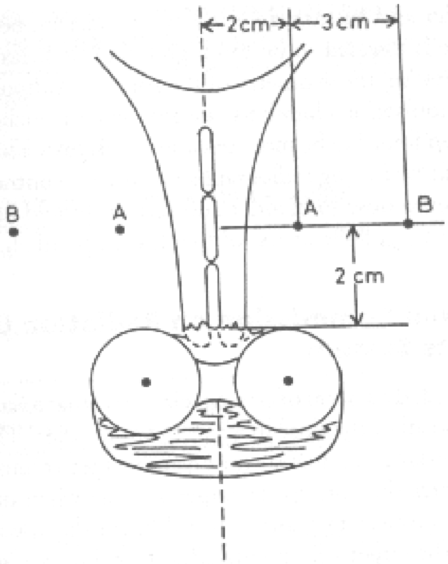


FIG. 15.23. Original definition of points *A* and *B*, according to the Manchester system. (From Meredith WJ. *Radium dosage: the Manchester system*. Edinburgh: Livingstone, 1967, with permission.)

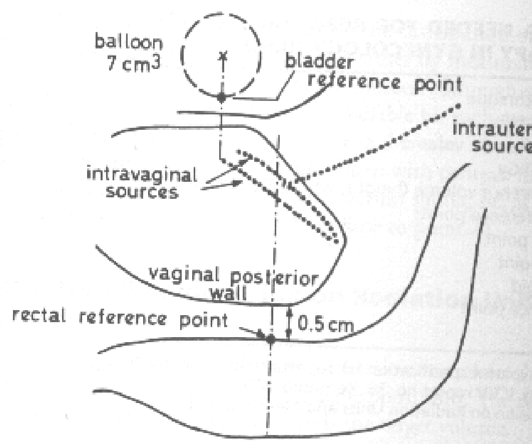


FIG. 15.27. Localization of bladder and rectum points. (From ICRU. Dose and volume specification for reporting intracavitary therapy in gynecology. ICRU Report No. 38. Bethesda, MD: International Commission on Radiation Units and Measurements, 1985, with permission.)

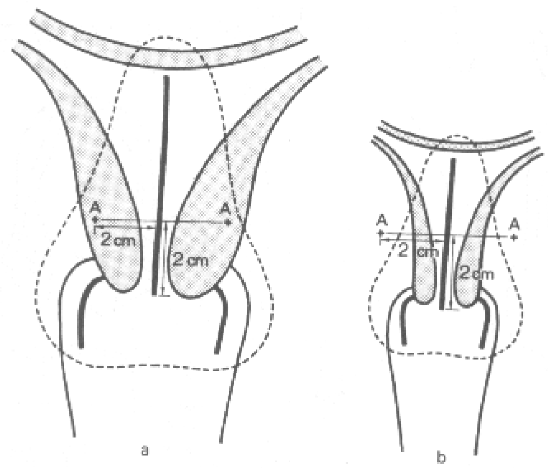


FIG. 15.24. Variation of point *A* relative to anatomy. **A:** Point *A* inside large cervix, resulting in underdosage. **B:** Point *A* outside small cervix, resulting in overdosage. (From Pierquin B, Wilson JF, Chassagne D, eds. *Modern brachytherapy.* New York: Masson, 1987, with permission.)

Data needed for reporting Intracavitary Therapy in Gynecology (ICRU)
- Description of Technique
- Total reference air kerma
- Description of the reference volume
  - Dose level if not 60 Gy
  - Dimensions of reference volume (height, width, thickness)
- Absorbed dose at reference points
  - Bladder reference point
  - Rectal reference point
  - Lymphatic trapezoid
  - Pelvic wall reference point
- Time-Dose pattern

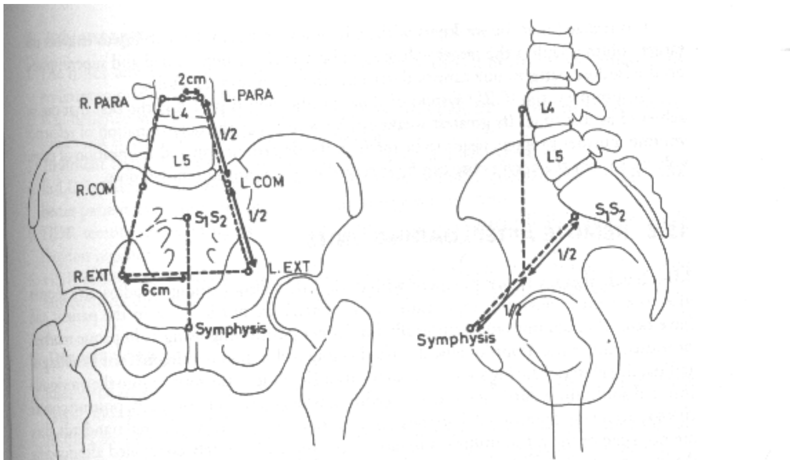


FIG. 15.28. Determination of reference points corresponding to the lymphatic trapezoid of Fletcher. (From ICRU. Dose and volume specification for reporting intracavitary therapy in gynecology. ICRU Report No. 38. Bethesda, MD: International Commission on Radiation Units and Measurements, 1985, with permission.)

- Lymphatic Trapezoid of Fletcher: corresponds to the paraaortic and iliac nodes

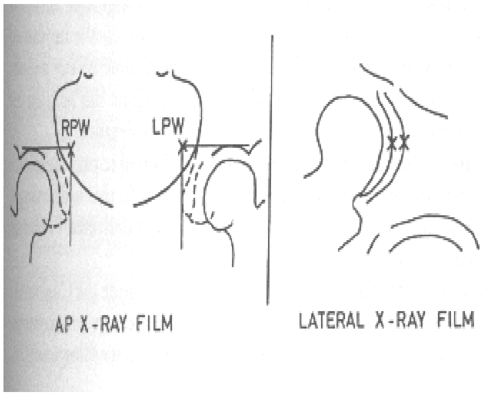


FIG. 15.29. Definition of pelvic wall points. *Left,* Anteroposterior view. *Right,* Lateral view. (From ICRU. Dose and volume specification for reporting intracavitary therapy in gynecology. ICRU Report No. 38. Bethesda, MD: International Commission on Radiation Units and Measurements, 1985, with permission.)

- Pelvic wall points
  - AP film: intersection of tangent to superior aspect of acetabulum and vertical line touching medial aspect of acetabulum
  - Lateral film: highest mid-distance pts of the left and right acetabulums

Intracavitary: Vaginal and Endometrial

Typically treated with a vaginal cylinder. No consistent prescription point, but prescribing to vaginal surface or to 5 mm depth are most common.

Miscellaneous

HDR: > 12 Gy/hour
LDR: < 2 Gy/hour

$1/T_{eff} = 1/T_{phys} + 1/T_{biol}$

Chapter 16---Radiation Protection

The NRC has control over all reactor-produced materials, while naturally occurring radioactive materials and x-ray machines are regulated by individual states.

16.1 Dose Equivalent

= Absorbed dose x Quality Factor
- Quality factor is analogous to RBE

SI Unit = Sievert = 1 J/kg
- 100 rem = 1 Sievert

16.3 Background Radiation

From 3 sources: terrestial radiation, cosmic radiation, and radiation from radioactive elements in our bodies
- avg total background is 3 mSv/yr
  - Radon accounts for 2 mSv/yr

Radiation exposure also from medical procedures.

Exposures from natural background radiation and medical procedures are not included in the occupational exposure controls for individual cases.

16.4 Low Level Radiation Effects

Stochastic effect: probability increases with dose, but severity does not.

Deterministic effect: Severity increases with increasing dose.

For purpose of radiation protection: assume a linear without threshold model

16.5 Effective Dose Equivalent Limits

TABLE 16.5. SUMMARY OF RECOMMENDATIONS

A. Occupational exposures (annual)		
1. Effective dose equivalent limit (stochastic effects)	50 mSv	(5 rem)
2. Dose equivalent limits for tissues and organs (nonstochastic effects)		
a. Lens of eye	150 mSv	(15 rem)
b. All others (e.g., red bone marrow, breast, lung, gonads, skin, and extremities)	500 mSv	(50 rem)
3. Guidance: cumulative exposure	10 mSv × age	(1 rem × age in yr)
B. Planned special occupational exposure, effective dose equivalent limit	see section 15[a]	
C. Guidance for emergency occupational exposure	See section 16[a]	
D. Public exposures (annual)		
1. Effective dose equivalent limit, continuous or frequent exposure	1 mSv	(0.1 rem)
2. Effective dose equivalent limit, infrequent exposure	5 mSv	(0.5 rem)
3. Remedial action recommended when:		
a. Effective dose equivalent	>5 mSv	(>0.5 rem)
b. Exposure to radon and its decay products	>0.007 Jhm^{-3}	(>2 WLM)
4. Dose equivalent limits for lens of eye, skin, and extremities	50 mSv	(5 rem)
E. Education and training exposures (annual)		
1. Effective dose equivalent	1 mSv	(0.1 rem)
2. Dose equivalent limit for lens of eye, skin, and extremities	50 mSv	(5 rem)
F. Embryo-fetus exposures		
1. Total dose equivalent limit	5 mSv	(0.5 rem)
2. Dose equivalent limit in a month	0.5 mSv	(0.05 rem)
G. Negligible individual risk level (annual) effective dose equivalent per source or practice	0.01 mSv	(0.001 rem)

[a]In NCRP Report no. 91.
From NCRP. Recommendations on limits for exposure to ionizing radiation. Report no. 91. Bethesda, MD: National Council on Radiation Protection and Measurements, 1987, with permission.

16.6 Structural Shielding Design

For protection calculations, the dose equivalent limit is 0.1 rem/week, which is equivalent to 5 rem/yr.

Protection required vs. 3 types of radiation: primary radiation, scatter, and leakage radiation through source housing.
- Primary barrier: barrier for the useful beam
- Secondary barrier: barrier for leakage and scatter

Following factors impt when considering barrier thickness:
1) *Workload (W):* Expressed in weekly dose delivered at 1 m from source (expressed in rad/week at 1 m)
2) *Use Factor (U):* Fraction of operating time during which radiation is directed toward a particular barrier.
3) *Occupancy Factor (T):* Fraction of operating time during which area of interest is occupied by the individual.
    a. Occupancy factor always = 1 in controlled area
    b. For low occupancy areas, occupancy factors are fractional
    c. Controlled areas have T = 1, allowed 100 mrem/week, or 5 rem/year
        i. In practice, controlled areas shielded to limit dose rate to < 2 mrem in any hour
    d. Non-controlled areas: required to be shielded to 2 mrem/week, or 100 mrem/year
4) *Distance (d):* in meters from source to the area to be protected.

For Primary Radiation Barrier: if the maximum permissible dose equivalent for the area to be protected is P (0.1 rad/week for controlled, and 0.01 rad/week for noncontrolled areas), and B is the transimission factor for the barrier to reduce primary beam dose to P in the area of interest, then:

$$P = WUTB / d2$$

Door Shielding: A maze arrangement, by preventing direct incidence of radiation at the door, dramatically reduces the thickness of shielding required. With a maze, the door is primarily exposed to scatter. For megavoltage beams, scattered beams have an energy of 500 kVp or less. In most cases, required door shielding is 6 mm of lead or less.

Protection against Neutrons: X-ray beams > 10 MV are contaminated with neutrons, produced when photons interact with the materials in the target, flattening filter, collimator, and other shielding components. The door must be protected against neutrons that diffuse into the maze and reach the door. Neutron fluence at door is 1% of that seen at the machine. A longer maze (>5m) is preferable to further reduce neutron fluence at door. Shielding accomplished by hydrogenous material (polyethylene) in the door.

16.7 Protection Against Radiation from Brachytherapy Sources

Storage: Lead-lined safe with lead-filled drawers. Storage area for radium should be ventilated and filtered to outdoors, to prevent it being drawn up into the general ventilation system.

Source Preparation: Source preparation bench should be provided close to the safe. Personnel should work behind barriers to shield (lead glass viewing window)

Source Transportation: sources transported in lead containers

Leak Testing: Various methods of leak testing are available, and periodic tests are specified by state regulations.

16.8 Radiation Protection Surveys

Radiation Monitoring Instruments: Detectors most often used for x-ray measurements are ionization chambers, Geiger counters, TLD's, and film (TLD and film discussed previously)

- Ionization Chamber: those used for low-level x-ray measurements have a large volume to obtain high sensitivity
- Geiger-Muller Counters: Much more sensitive than ionization chamber, but it is not a dose-measuring device. Although useful for preliminary surveys to detect presence of radiation, ion chambers are better for quantitative measurements. Portable, highly sensitive, instantaneous readout. Good for locating a dropped I-125 seed
- Neutron Detectors: often contain filling gases such as BF_3

Equipment Survey

- Leakage radiation can be measured by using an ion chamber, with dose rate determined at a distance of 1 m from the source. Measurements should be made in direction in which leakage is expected to be greatest
- For Co-60, leakage radiation in the off position is measured at 14 different directions to determine avg and max leakage dose rate. An ion chamber such as a Cutie Pie is useful for this

Area Survey:

- areas outside the trt room accessible to individuals need to be surveyed, and should be designated as controlled or noncontrolled, depending on whether exposure of persons in the area is monitored or not.
- Should take into account workload, use factor, occupancy factor

16.9 Personnel Monitoring

Used in controlled areas for occupationally exposed individuals. Mostly performed with film badges, which monitors whole body exposure (worn on chest or abdomen).

- Drawbacks to film badges
    - Energy dependence is a problem

Pocket dosimeters can be used in situations where exposure needs to be monitored more frequently than possible with film badge.

16.10 NRC Regulations

- Patient administered radiopharmaceutical: cannot be released from confinement until measured dose rate at 1m is < 5 mrem/h or activity remaining in pt is < 30 microCurie
- Permanent implant: can't release pt until dose 1m from pt is < 5 mrem/h
- If a misadministration occurs, NRC and an M.D. must be notified within 24 hours

Chapter 17---Quality Assurance

17.4 Dosimetric Accuracy: Need for accuracy of +/- 5% in dose delivery.

Multiple Beam Alignment Check: To check alignment between multiple beams, can perform the split-field test.

- double expose a film to two fields, spaced 180 degrees apart
    - Check alignment of the exposures

Field Flatness: Flatness criteria should reflect the effect of the flattening filter, and not the penumbra. Should be within +/- 3%.

TABLE 17.8. PERIODIC QA OF LINEAR ACCELERATORS

Frequency	Procedure	Tolerance[a]
Daily	Dosimetry	
	X-ray output constancy	3%
	Electron output constancy[b]	3%
	Mechanical	
	Localizing lasers	2 mm
	Distance indicator (ODI)	2 mm
	Safety	
	Door interlock	Functional
	Audiovisual monitor	Functional
Monthly	Dosimetry	
	x-ray output constancy[c]	2%
	Electron output constancy[c]	2%
	Backup monitor constancy	2%
	x-ray central axis dosimetry parameter (PDD, TAR) constancy	2%
	Electron central axis dosimetry parameter constancy (PDD)	2 mm at therapeutic depth
	x-ray beam flatness constancy	2%
	Electron beam flatness constancy	3%
	x-ray and electron symmetry	3%
	Safety interlocks	
	Emergency off switches	Functional
	Wedge, electron cone interlocks	Functional
	Mechanical checks	
	Light/radiation field coincidence	2 mm or 1% on a side[d]
	Gantry/collimator angle indicators	1°
	Wedge position	2 mm (or 2% change in transmission factor)
	Tray position	2 mm
	Applicator position	2 mm
	Field size indicators	2 mm
	Cross-hair centering	2 mm diameter
	Treatment couch position indicators	2 mm/1°
	Latching of wedges, blocking tray	Functional
	Jaw symmetry[e]	2 mm
	Field light intensity	Functional
Annual	Dosimetry	
	x-ray/electron output calibration constancy	2%
	Field size dependence of x-ray output constancy	2%
	Output factor constancy for electron applicators	2%
	Central axis parameter constancy (PDD, TAR)	2%
	Off-axis factor constancy	2%
	Transmission factor constancy for all treatment accessories	2%
	Wedge transmission factor constancy[f]	2%
	Monitor chamber linearity	1%
	x-ray output constancy vs gantry angle	2%
	Electron output constancy vs gantry angle	2%
	Off-axis factor constancy vs gantry angle	2%
	Arc mode	Mfrs. specs.
	Safety interlocks	
	Follow manufacturers test procedures	Functional
	Mechanical checks	
	Collimator rotation isocenter	2 mm diameter
	Gantry rotation isocenter	2 mm diameter
	Couch rotation isocenter	2 mm diameter
	Coincidence of collimetry, gantry, couch axes with isocenter	2 mm diameter
	Coincidence of radiation and mechanical isocenter	2 mm diameter
	Table top sag	2 mm
	Vertical travel of table	2 mm

[a]The tolerances listed in the tables should be interpreted to mean that if a parameter either: (1) exceeds the tabulated value (e.g., the measured isocenter under gantry rotation exceeds 2 mm diameter); or (2) that the change in the parameter exceeds the nominal value (e.g., the output changes by more than 2%), then an action is required. The distinction is emphasized by the use of the term constancy for the latter case. Moreover, for constancy, percent values are ± the deviation of the parameter with respect its nominal value; distances are referenced to the isocenter or nominal SSD.
[b]All electron energies need not be checked daily, but all electron energies are to be checked at least twice weekly.
[c]A constancy check with a field instrument using temperature/pressure corrections.
[d]Whichever is greater. Should also be checked after change in light field source.
[e]Jaw symmetry is defined as difference in distance of each jaw from the isocenter.
[f]Most wedges' transmission factors are field size and depth dependent.
From AAPM. Comprehensive QA for radiation oncology: report of the AAPM Radiation Therapy Committee Task Group 40. *Med Phys* 1994;21:581–618, with permission.

Chapter 18---Total Body Irradiation

Techniques and Equipment

Several techniques are used. Among most common are APPA and opposed lats.

1) APPA
- Better dose uniformity along longitudinal axis
- Positioning is more problematic vs. opposed lats
- Pt treated upright
  - Standing TBI allows shielding of critical organs from photons and boosting of superficial tissues in the shadow of the blocks with electrons
    - Example: lung dose reduced using lung blocks of 1 HVL and chest wall under blocks can be boosted with electrons

2) Opposed lats
- Position: pt seated (semi-fetal position) or supine
- Greater variation in body thickness compared to APPA
  - Compensator are used for HN, lungs, and legs to achieve a dose uniformity of +/- 10%
    - Compensators are mounted in the treatment head
  - The reference thickness for compensation is the lateral diameter of the body at the level of the umbilicus
- Arms placed at side to increase shielding for lungs

Beam Energy

- If pt > 35 cm thick, use > 6 MV photons to avoid tissue lateral effect (excessive hot spot at dmax compared to midplane)
- If pt < 35 cm thick, can use 6 MV photons

Initial Dose Build-up
- Bolus or beam spoiler is specified to bring the surface dose to at least 90% of the prescribed TBI dose.
  - Large spoiler screen of 1-2 cm thick acrylic is enough to meet these requirements
    - Screen placed as close as possible to patient surface

Field Size
- Although patient dimensions vary, scatter factors are not too sensitive to field size variation for large fields.
- Reasonable to use fixed equivalent field size
  - 40 x 40 cm field for large pts, 30 x 30 cm for peds patients

Compensator Design

Most TBI protocols require dose homogeneity along the body axis to be within 10%. This requirement cannot be met without the use of compensators. The thickness of compensator required depends on:
- tissue deficit compared to the reference depth at the prescription point
- density of the compensator (Aluminium is commonly used)
- distance of the compensator from the point of dose compensation
- depth of the point of dose compensation
- field size
- beam energy

Thickness ratio: required thickness of tissue-equivalent compensator that give the same dose at the point of interest as would a bolus of thickness equal to the tissue deficit.
- this ratio = 0.70 for most beam energies and compensation conditions

In Vivo Patient Doseimetry

TLD results should be compared with expected doses.
- agreement of +5% between calculated and measured doses is considered good
- Overall dose uniformity of +10% is acceptable for most protocols

Chapter 19---Three-dimensional Conformal Radiation Therapy

Dose Volume Histograms:
- Cumulative DVH: plot of the volume of a given structure receiving a certain dose or higher as a function of dose
- Differential DVH: plot of volume receiving a dose within a specified dose interval as a function of dose

Dose Computation Algorithms:
- Should have acceptable accuracy
  - +/- 3% for homogeneous and +/- 5% for heterogeneous tissues
- 3-categories of algorithms:
  - Correction-based (accuracy limited for 3-D heterogeneity corrections in lung and tissue interfaces, especially in situations where electronic equilibrium is not fully established)
  - Model-based (Convolution-superposition)
  - Monte-Carlo (most accurate method of calculating dose distribution)

Chapter 20---Intensity-Modulated Radiation Therapy

Dynamic MLC IMRT

- If the gradient of the intensity profile is positive (increasing fluence), the leading leaf should move at maximum speed and the trailing leaf should provide the required intensity modulation
- If the spatial gradient of the intensity profile is negative (decreasing fluence), the trailing leaf should move at the max speed, and the leading leaf should provide the required intensity modulation.

Multileaf Collimator Transmission

- Interleaf transmission with the "tongue and groove" is about 2.5%

Head Scatter

- If the MLC in the linac head is installed closer to the patient surface than the collimator jaws, the Sc factor depends mostly on the jaw opening and not on the MLC opening
- If the MLC is located above the collimator jaws, the head scatter would be affected more by the MLC setting than the jaw opening.

Quality Assurance

Doses should be accurate to $\pm$ 3%

Chapter 21---Stereotactic Radiosurgery

Best attainable mechanical accuracy of isocenter displacement from defined isocenter is 0.2 mm $\pm$ 0.1 mm, although a maximum error of $\pm$ 1 mm is commonly accepted for uncertainty in target localization.

Three types of radiation used: heavy-charged particles, Co-60 gamma rays, and megavoltage x-rays

Linac-based SRS

- consists of using multiple noncoplanar arcs of circular (dynamically shaped) beams converging on to the machine isocenter
- pedestal-mounted stereotactic frames are more accurate than couch-mounted frames
- geometric accuracy of target localization for MRI is not as good as for CT or angiography. Often use multiple imaging modalities for improved target localization

Gamma Knife

- 201 Co-60 sources. Collimated to focus on a single point at a source to focus distance of 40.3 cm
- Circular field opening max of 18 mm at the focus point

- Therefore, can be used only for small lesions because of field size limitation. However, several isocenters can be placed within the same target to expand dose distribution
- For treating multiple isocenters, gamma knife is more practical because of simplicity of setup
- Gamma knife can produce a more conformal dose distribution than that possible with Linac unless Linac is equipped with dynamic MLC
    - However, Linac based SRS gives more homogenous dose, b/c gamma-knife often prescribed to the 50% IDL

Chapter 22---High Dose Rate Brachytherapy

- HDR: defined as delivering > 12 Gy/hr
- Iridium-192 most commonly used radioisotope
    - Pros:
        - High specific activity (smaller source for same activity)
        - Lower photon energy (requires less shielding)
    - Cons:
        - Shorter half life (73.8 days), requiring source replacement q3-4 mos
- QA details
    - Source calibration should be within 5% of manufacturer
    - Source position calibration within 1 mm
- Strength of brachytherapy source: specified in air kerma strength
    - Air kerma strength is determined from exposure rate measured in free air at a distance of 1 m from the source

Chapter 23---Prostate Implants

- I-125 or Pd-103 used
    - Dose rate for Pd-103 is 3x that for I-125
    - Pd-103 has shorter half-life, so 70% of prescribed dose is delivered in first month
- Transperineal approach
- Only 10% of all loose seeds should be assayed

Treatment planning:

- Major problem is usual disagreement between pre- and post-implant dose distributions
    - Hot and cold spots can develop as a result of source movement with time
- Another problem: source anisotropy
    - Cold spots of greater than 50% can exist at the ends
    - More of a problem if sources are aligned permanently end to end with each other along straight lines
    - Randomness that develops after implantation reduces the overall anisotropy effect
- Dose computation: TG-43 used
- Calibration: Agreement within $\pm$ 5% with the vendor calibration is acceptable