Chapter 8
Electron Beams: Physical and Clinical Aspects

This set of 91 slides is based on Chapter 8 authored by W. Strydom, W. Parker, and M. Olivares of the IAEA publication (ISBN 92-0-107304-6):

Radiation Oncology Physics:
A Handbook for Teachers and Students

Objective:

To familiarize students with basic physical and clinical principles of radiotherapy with megavoltage electron beams.

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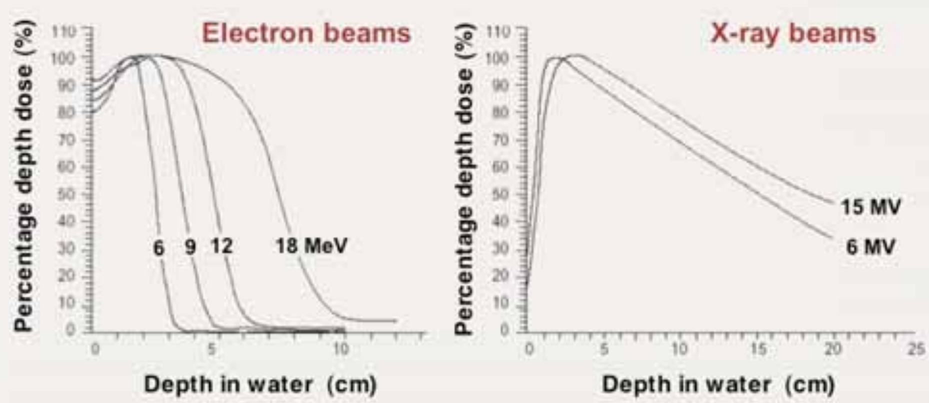
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS

- Megavoltage electron beams represent an important treatment modality in modern radiotherapy, often providing a unique option in the treatment of superficial tumours.
  - Electrons have been used in radiotherapy since the early 1950s.
  - Modern high-energy linacs typically provide, in addition to two photon energies, several electron beam energies in the range from 4 MeV to 25 MeV.

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS

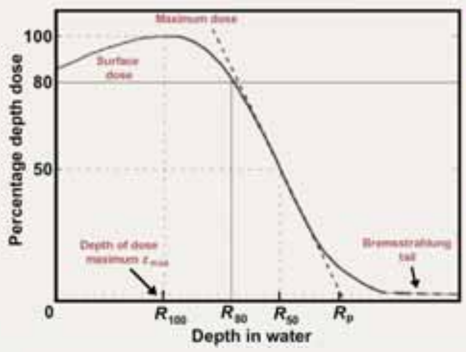
8.1.1 General shape of the depth dose curve

- The general shape of the central axis depth dose curve for electron beams differs from that of photon beams.



8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.1 General shape of the depth dose curve

- The electron beam central axis percentage depth dose curve exhibits the following characteristics:



- The surface dose is relatively high (of the order of 80 - 100%).
- Maximum dose occurs at a depth referred to as the depth of dose maximum z_{max}.
- Beyond z_{max} the dose drops off rapidly and levels off at a small low level dose called the brems-strahlung tail (of the order of a few per cent).

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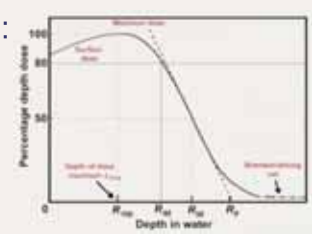
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.1 General shape of the depth dose curve

- Electron beams are almost monoenergetic as they leave the linac accelerating waveguide.
- In moving toward the patient through:
  - Waveguide exit window
  - Scattering foils
  - Transmission ionization chamber
  - Air



and interacting with photon collimators, electron cones (applicators) and the patient, bremsstrahlung radiation is produced. This radiation constitutes the bremsstrahlung tail of the electron beam PDD curve.

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8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.2 Electron interactions with absorbing medium

- As the electrons propagate through an absorbing medium, they interact with atoms of the absorbing medium by a variety of elastic or inelastic Coulomb force interactions.
- These Coulomb interactions are classified as follows:
  - Inelastic collisions with orbital electrons of the absorber atoms.
  - Inelastic collisions with nuclei of the absorber atoms.
  - Elastic collisions with orbital electrons of the absorber atoms.
  - Elastic collisions with nuclei of the absorber atoms.

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.2 Electron interactions with absorbing medium

- Inelastic collisions between the incident electron and orbital electrons of absorber atoms result in loss of incident electron's kinetic energy through ionization and excitation of absorber atoms (collision or ionization loss).
- Absorber atoms can be ionized through two types of ionization collision:
  - Hard collision in which the ejected orbital electron gains enough energy to be able to ionize atoms on its own (these electrons are called delta rays).
  - Soft collision in which the ejected orbital electron gains an insufficient amount of energy to be able to ionize matter on its own.

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.2 Electron interactions with absorbing medium

- Elastic collisions between the incident electron and nuclei of the absorber atoms result in:
  - Change in direction of motion of the incident electron (elastic scattering).
  - A very small energy loss by the incident electron in individual interaction, just sufficient to produce a deflection of electron's path.
- Incident electron loses kinetic energy through cumulative action of multiple scattering events, each scattering event characterized by a small energy loss.

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8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.2 Electron interactions with absorbing medium

- Electrons traversing an absorber lose their kinetic energy through ionization collisions and radiation collisions.
- The rate of energy loss per gram and per cm^2 is called the mass stopping power and it is a sum of two components:
  - Mass collision stopping power
  - Mass radiation stopping power
- The rate of energy loss for a therapy electron beam in water and water-like tissues, averaged over the electron's range, is about 2 MeV/cm.

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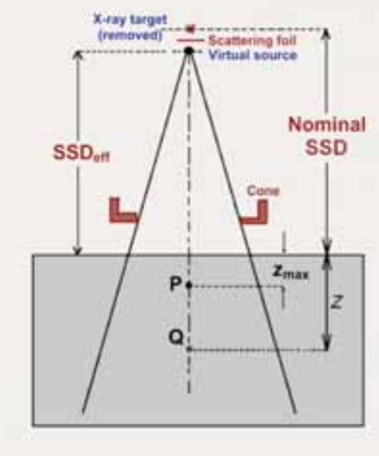
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.3 Inverse square law (virtual source position)

- In contrast to a photon beam, which has a distinct focus located at the accelerator x ray target, an electron beam appears to originate from a point in space that does not coincide with the scattering foil or the accelerator exit window.
- The term "virtual source position" was introduced to indicate the virtual location of the electron source.



8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.3 Inverse square law (virtual source position)

- Effective source-surface distance SSD$_{eff}$ is defined as the distance from the virtual source position to the edge of the electron cone applicator.
- The inverse square law may be used for small SSD differences from the nominal SSD to make corrections to absorbed dose rate at z_{max} in the patient for variations in air gaps g between the actual patient surface and the nominal SSD.

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.3 Inverse square law (virtual source position)

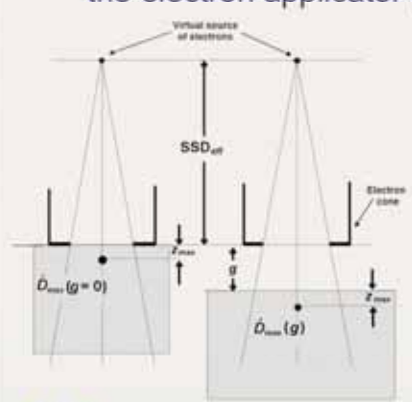
- A common method for determining SSD_{eff} consists of measuring the dose rate at z_{max} in phantom for various air gaps g starting with $\dot{D}_{max}(g=0)$ at the electron cone.
  - The following inverse square law relationship holds:

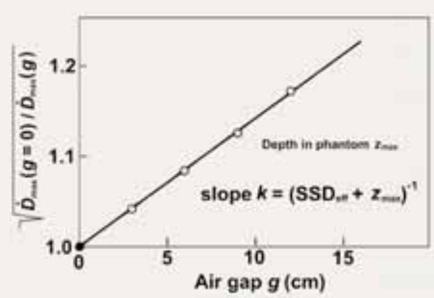
$$\frac{\dot{D}_{max}(g=0)}{\dot{D}_{max}(g)} = \left\{\frac{SSD_{eff} + z_{max} + g}{SSD_{eff} + z_{max}}\right\}^2$$

  - The measured slope of the linear plot is: $k = \frac{1}{SSD_{eff} + z_{max}}$
  - The effective SSD is then calculated from: $SSD_{eff} = \frac{1}{k} + z_{max}$

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.3 Inverse square law (virtual source position)

- Typical example of data measured in determination of virtual source position SSD_{eff} normalized to the edge of the electron applicator (cone).





$$SSD_{eff} = \frac{1}{k} + z_{max}$$

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.3 Inverse square law (virtual source position)

- For practical reasons the nominal SSD is usually a fixed distance (e.g., 5 cm) from the distal edge of the electron cone (applicator) and coincides with the linac isocentre.
- Although the effective SSD (i.e., the virtual electron source position) is determined from measurements at z_{max} in phantom, its value does not change with change in the depth of measurement.
- The effective SSD depends on electron beam energy and must be measured for all energies available in the clinic.

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8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

- By virtue of being surrounded by a Coulomb force field, charged particles, as they penetrate into an absorber encounter numerous Coulomb interactions with orbital electrons and nuclei of the absorber atoms.
  - Eventually, a charged particle will lose all of its kinetic energy and come to rest at a certain depth in the absorbing medium called the particle range.
  - Since the stopping of particles in an absorber is a statistical process several definitions of the range are possible.

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Definitions of particle range: (1) CSDA range

- In most encounters between the charged particle and absorber atoms the energy loss by the charged particle is minute so that it is convenient to think of the charged particle as losing its kinetic energy gradually and continuously in a process referred to as the continuous slowing down approximation (CSDA - Berger and Seltzer).
- The CSDA range or the mean path length of an electron of initial kinetic energy E_o can be found by integrating the reciprocal of the total mass stopping power over the energy from E_o to 0:

$$R_{\mathrm{CSDA}} = \int_0^{E_o} \left[\frac{S(E)}{\rho} \right]^{-1} \mathrm{d}E$$

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

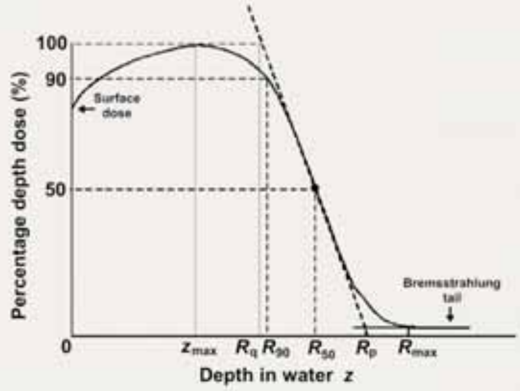
CSDA range for electrons in air and water

Electron energy (MeV)	CSDA range in air (g/cm^2)	CSDA range in water (g/cm^2)
6	3.255	3.052
7	3.756	3.545
8	4.246	4.030
9	4.724	4.506
10	5.192	4.975
20	9.447	9.320
30	13.150	13.170

- CSDA range is a calculated quantity that represents the mean path length along the electron's trajectory.
- CSDA range is not the depth of penetration along a defined direction.

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

- Several other range definitions are in use for electron beams:
  - Maximum range R_{max}
  - Practical range R_p
  - Therapeutic range R_{90}
  - Therapeutic range R_{80}
  - Depth R_{50}
  - Depth R_q



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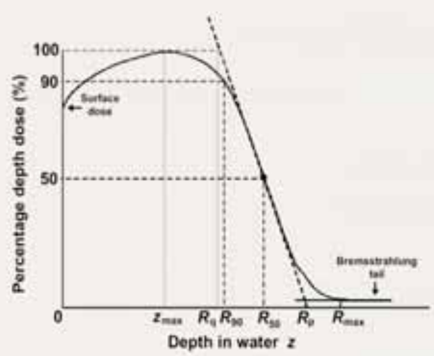
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

- Maximum range R_{max} is defined as the depth at which the extrapolation of the tail of the central axis depth dose curve meets the bremsstrahlung background.

  R_{max} is the largest penetration depth of electrons in absorbing medium.



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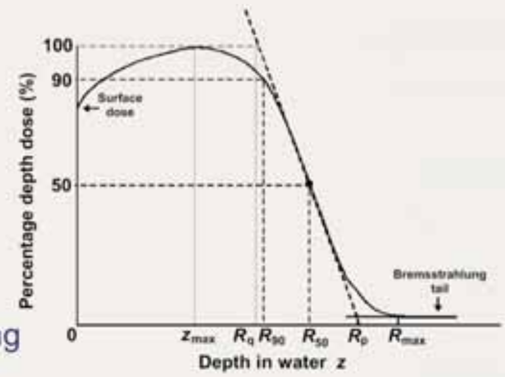
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

- Practical range R_p
  is defined as the depth at which the tangent plotted through the steepest section of the electron depth dose curve intersects with the extrapolation line of the bremsstrahlung tail.



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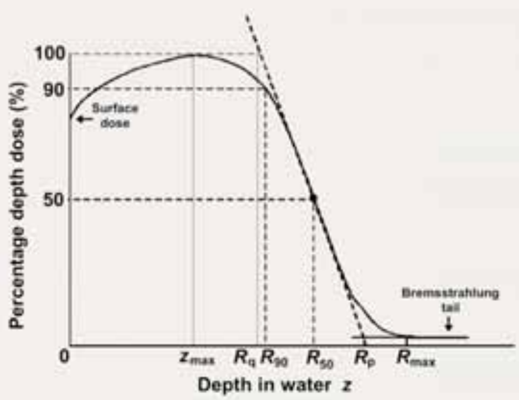
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

- Depths R_{90}, R_{80}, and R_{50} are defined as depths on the electron PDD curve at which the PDDs beyond the depth of dose maximum z_{max} attain values of 90%, 80%, and 50%, respectively.



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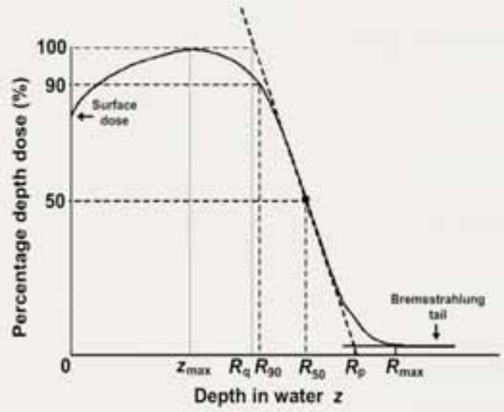
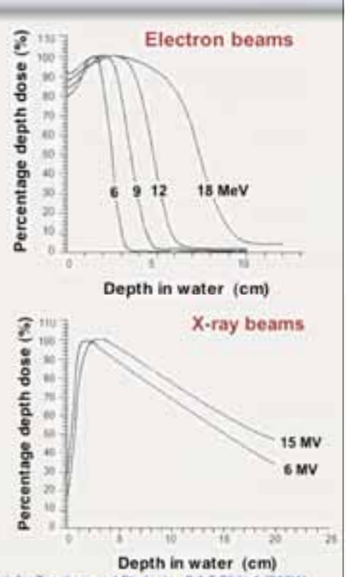
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.4 Range concept

- Depth R_q is defined as that depth where the tangent through the dose inflection point intersects the maximum dose level.



8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.5 Buildup region

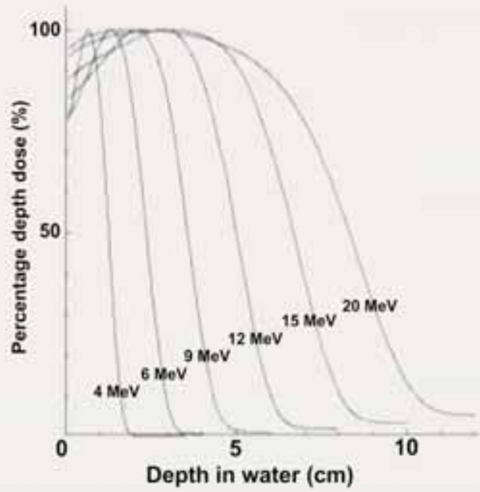
- The buildup region for electron beams, like for photon beams, is the depth region between the phantom surface and the depth of dose maximum z_{max}.
- The surface dose for megavoltage electron beams is relatively large (typically between 75% and 95%) in contrast to the surface dose for megavoltage photon beams which is of the order of 10% to 25%.



8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS

8.1.5 Buildup region

- Unlike in photon beams, the percentage surface dose in electron beams increases with increasing energy.
- In contrast to photon beams, z_{max} in electron beams does not follow a specific trend with electron beam energy; it is a result of machine design and accessories used.



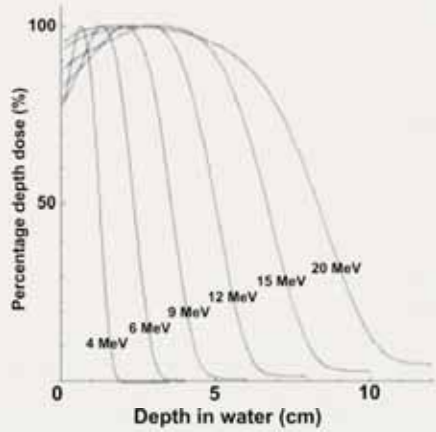
8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS

8.1.6 Dose distribution beyond z_{max}

- The dose beyond z_{max}, especially at relatively low megavoltage electron beam energies, drops off sharply as a result of the scattering and continuous energy loss by the incident electrons.
- As a result of bremsstrahlung energy loss by the incident electrons in the head of the linac, air and the patient, the depth dose curve beyond the range of electrons is attributed to the bremsstrahlung photons.

8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.6 Dose distribution beyond z_{max}

- The bremsstrahlung contamination of electron beams depends on electron beam energy and is typically:
  - Less than 1% for 4 MeV electron beams.
  - Less than 2.5% for 10 MeV electron beams.
  - Less than 4% for 20 MeV electron beams.

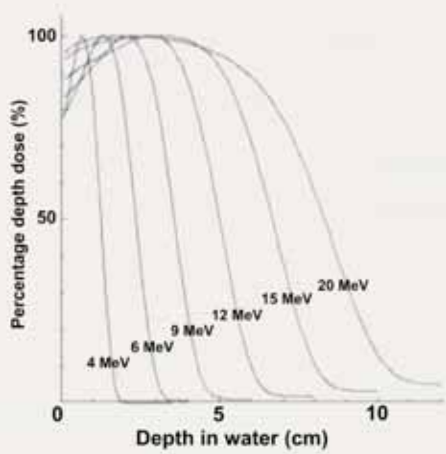


8.1 CENTRAL AXIS DEPTH DOSE DISTRIBUTIONS
8.1.6 Dose distribution beyond z_{max}

- Electron dose gradient G is defined as follows:

$$G = \frac{R_p}{R_p - R_q}$$

- The dose gradient G for lower electron beam energies is steeper than that for higher electron energies.

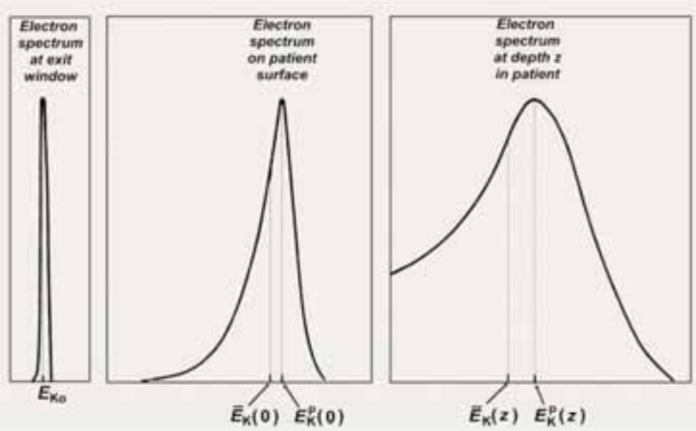


8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.1 Electron beam energy specification

- The spectrum of the electron beam is very complex and is influenced by the medium the beam traverses.
  - Just before exiting the waveguide through the beryllium exit window the electron beam is almost monoenergetic.
  - The electron energy is degraded randomly when electrons pass through the exit window, scattering foil, transmission ionization chamber and air. This results in a relatively broad spectrum of electron energies on the patient surface.
  - As the electrons penetrate into tissue, their spectrum is broadened and degraded further in energy.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.1 Electron beam energy specification

- The spectrum of the electron beam depends on the point of measurement in the beam.



8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS

8.2.1 Electron beam energy specification

- Several parameters are used for describing the beam quality of an electron beam:
  - Most probable energy $E_K^p(0)$ of the electron beam on phantom surface.
  - Mean energy $\bar{E}_K(0)$ of electron beam on the phantom surface.
  - Half-value depth R_{50} on the percentage depth dose curve of the electron beam.
  - Practical range R_p of the electron beam.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS

8.2.1 Electron beam energy specification

- The most probable energy $E_K^p(0)$ on the phantom surface is defined by the position of the spectral peak.
- $E_K^p(0)$, the most probable of the electrons, is related to the practical range R_p (in cm) of the electron beam through the following polynomial equation:

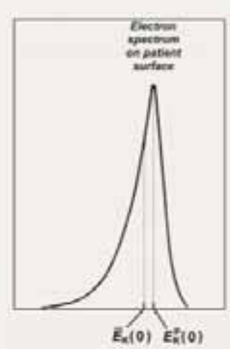
$$E_K^p(0) = C_1 + C_2 R_p + C_3 R_p^2$$

For water:

$C_1 = 0.22$ MeV

$C_2 = 1.98$ MeV/cm

$C_3 = 0.0025$ MeV/cm^2



8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.1 Electron beam energy specification

- The mean electron energy $\bar{E}_K(0)$ of the electron beam on the phantom surface is slightly smaller than the most probable energy $E_K^p(0)$ on the phantom surface as a result of an asymmetrical shape of the electron spectrum.

- The mean electron energy $\bar{E}_K(0)$ is related to the half-value depth R_{50} as:

$$\bar{E}_K(0) = CR_{50}$$

Constant C for water is 2.33 MeV/cm.

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8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.1 Electron beam energy specification

- Harder has shown that the most probable energy $E_K^p(z)$ and the mean energy $\bar{E}(z)$ of electron beam at a depth z in the phantom or patient decrease linearly with z.

- Harder's relationships are expressed as follows:

$$E_K^p(z) = E_K^p(0)\left\{1 - \frac{z}{R_p}\right\} \quad \text{and} \quad \bar{E}(z) \approx \bar{E}(0)\left\{1 - \frac{z}{R_p}\right\}$$

Note: $E_K^p(z=0) = E_K^p(0)$ $\qquad \bar{E}(z=0) = \bar{E}(0)$

$E_K^p(z=R_p) = 0$ $\qquad \bar{E}(z=R_p) = 0$

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8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS

8.2.2 Typical depth dose parameters as a function of energy

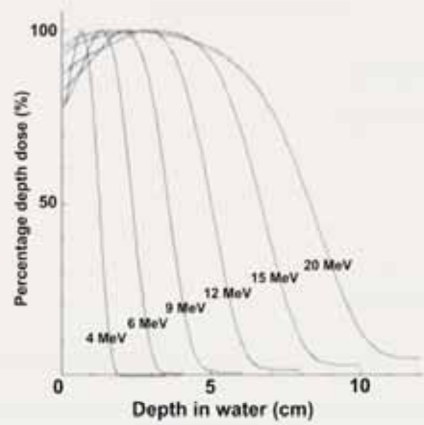
- Typical electron beam depth dose parameters that should be measured for each clinical electron beam.

Energy (MeV)	R_{90} (cm)	R_{80} (cm)	R_{50} (cm)	R_p (cm)	$\bar{E}(0)$ (MeV)	Surface dose %
6	1.7	1.8	2.2	2.9	5.6	81
8	2.4	2.6	3.0	4.0	7.2	83
10	3.1	3.3	3.9	4.8	9.2	86
12	3.7	4.1	4.8	6.0	11.3	90
15	4.7	5.2	6.1	7.5	14.0	92
18	5.5	5.9	7.3	9.1	17.4	96

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS

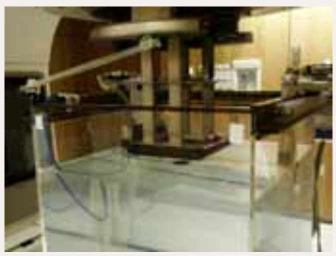
8.2.3 Percentage depth dose

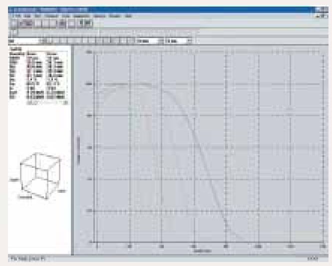
- Similarly to PDDs for photon beams, the PDDs for electron beams, at a given source-surface distance SSD, depend upon:
  - Depth z in phantom (patient).
  - Electron beam kinetic energy $E_K(0)$ on phantom surface.
  - Field size A on phantom surface.



8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

- The PDDs of electron beams are measured with:
  - Cylindrical, small-volume ionization chamber in water phantom.
  - Diode detector in water phantom.
  - Parallel-plate ionization chamber in water phantom.
  - Radiographic or radiochromic film in solid water phantom.





8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

- Measurement of electron beam PDDs:
  - If an ionization chamber is used, the measured depth ionization distribution must be converted into a depth dose distribution by using the appropriate stopping power ratios, water to air, at depths in phantom.
  - If diode is used, the diode ionization signal represents the dose directly, because the stopping power ratio, water to silicon, is essentially independent of electron energy and hence depth.
  - If film is used, the characteristic curve (H and D curve) for the given film should be used to determine the dose against the film density.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

Dependence of PDDs on electron beam field size.

- For relatively large field sizes the PDD distribution at a given electron beam energy is essentially independent of field size.
- When the side of the electron field is smaller than the practical range R_p, lateral electronic equilibrium will not exist on the beam central axis and both the PDDs as well as the output factors exhibit a significant dependence on field size.

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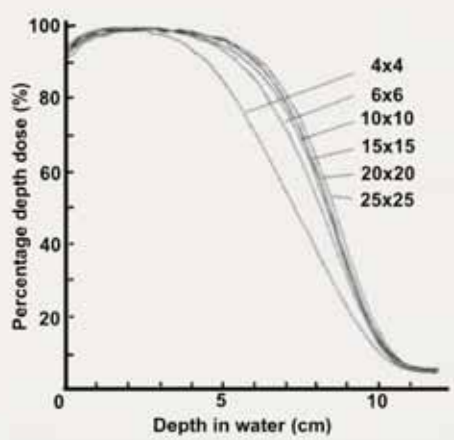
8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

PDDs for small electron fields

For a decreasing field size, when the side of the field decreases to below the R_p value for a given electron energy:

- Depth of dose maximum decreases.
- Surface dose increases.
- R_p remains essentially constant, except when the field size becomes very small.



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8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

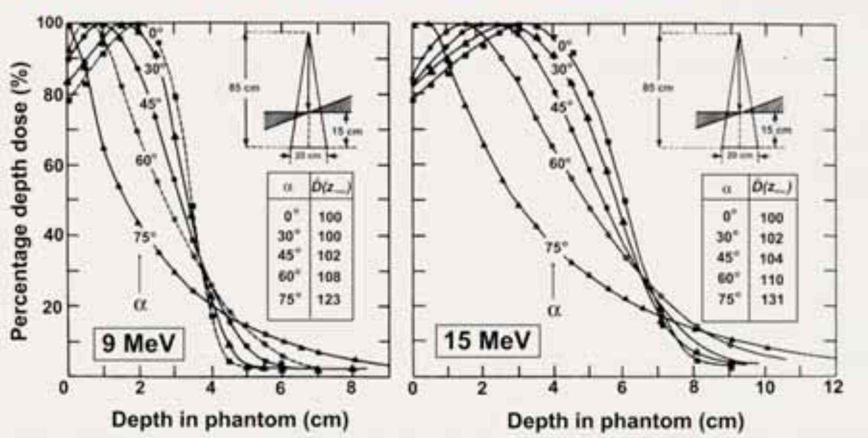
PDDs for oblique incidence.

- The angle of obliquity α is defined as the angle between the electron beam central axis and the normal to the phantom or patient surface. Angle $\alpha = 0$ corresponds to normal beam incidence.

- For oblique beam incidences, especially at large angles α the PDD characteristics of electron beams deviate significantly from those for normal beam incidence.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

- Percentage depth dose for oblique beam incidence



8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.3 Percentage depth dose

Depth dose for oblique beam incidence

- The obliquity effect becomes significant for angles of incidence α exceeding 45º.
- The obliquity factor OF(α,z) accounts for the change in depth dose at a given depth z in phantom and is normalized to 1.00 at z_{max} at normal incidence $\alpha = 0$.
- The obliquity factor at z_{max} is larger than 1 (see insets on previous slide).

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.4 Output factors

- The output factor
  - For a given electron energy and
  - For a given field size (delineated by applicator or cone)

  is defined as the ratio of the dose for the specific field size (applicator) to the dose for a 10x10 cm^2 reference field size (applicator), both measured:
  - At depth z_{max} on the beam central axis in phantom
  - At a nominal SSD of 100 cm.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.4 Output factors

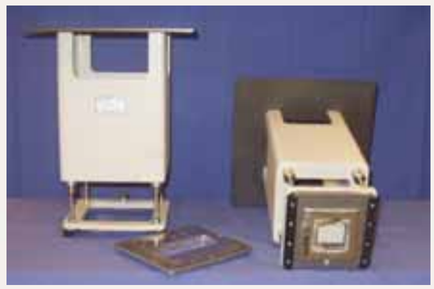
- When using electron beams from a linac, the photon collimator must be opened to the appropriate setting for a given electron applicator.
- Typical electron applicator sizes at nominal SSD are:
  - Circular with diameter: 5 cm
  - Square: 10x10 cm^2; 15x15 cm^2; 20x20 cm^2; and 25x25 cm^2.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.4 Output factors

- Often collimating blocks made of lead or a low melting point alloy (e.g., cerrobend) are used for field shaping. These blocks are attached to the end of the electron cone (applicator) and produce the required irregular field.
- Output factors, normalized to the standard 10x10 cm^2 electron cone, must be measured for all custom-made irregular fields.

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.4 Output factors

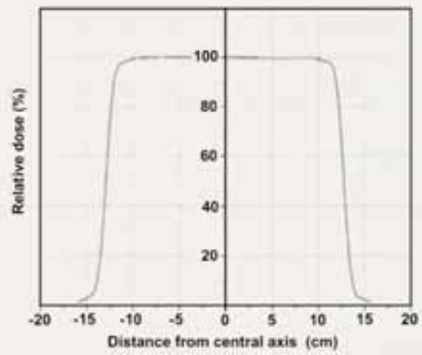
- For small irregular field sizes the extra shielding affects not only the output factors but also the PDD distribution because of the lack of lateral scatter.
- For custom-made small fields, in addition to output factors, the full electron beam PDD distribution should be measured.



8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.5 Therapeutic range

- The depth of the 90% dose level on the beam central axis (R_{90}) beyond z_{max} is defined as the therapeutic range for electron beam therapy.
  - R_{90} is approximately equal to $E_K/4$ in cm of water, where E_K is the nominal kinetic energy in MeV of the electron beam.
  - R_{80}, the depth that corresponds to the 80% PDD beyond z_{max}, may also be used as the therapeutic range and is approximated by $E_K/3$ in cm of water.

- A dose profile represents a plot of dose at a given depth in phantom against the distance from the beam central axis.
- The profile is measured in a plane perpendicular to the beam central axis at a given depth *z* in phantom.

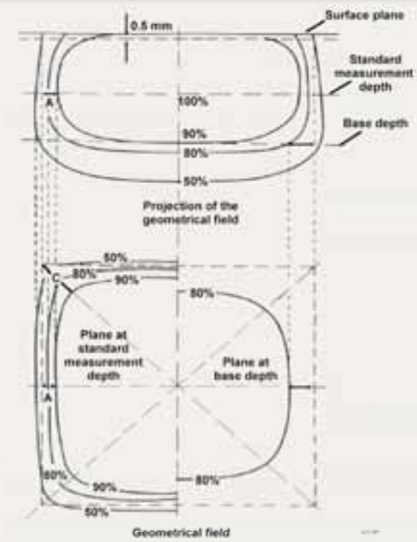


Dose profile measured at a depth of dose maximum z_{max} in water for a 12 MeV electron beam and 25x25 cm^2 applicator cone.

- Two different normalizations are used for beam profiles:
  - Profile data for a given depth in phantom may be normalized to the dose at z_{max} on the central axis (point P). The dose value on the beam central axis for $z \neq z_{max}$ then represents the central axis PDD value.
  - Profile data for a given depth in phantom may also be normalized to the value on the beam central axis (point Q). The values off the central axis for $z \neq z_{max}$ are then referred to as the off-axis ratios (OARs).

8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.7 Flatness and symmetry

- According to the International Electrotechnical Commission (IEC) the specification for beam flatness of electron beams is given for z_{max} under two conditions:
  - Distance between the 90% dose level and the geometrical beam edge should not exceed 10 mm along major field axes and 20 mm along diagonals.
  - Maximum value of the absorbed dose anywhere within the region bounded by the 90% isodose contour should not exceed 1.05 times the absorbed dose on the axis of the beam at the same depth.



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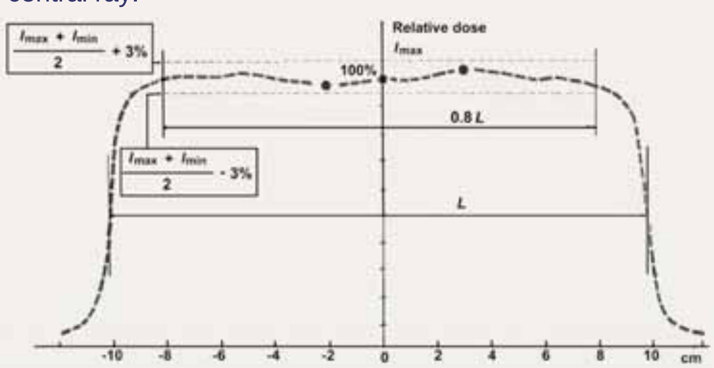
8.2 DOSIMETRIC PARAMETERS OF ELECTRON BEAMS
8.2.7 Flatness and symmetry

- According to the International Electrotechnical Commission (IEC) the specification for symmetry of electron beams requires that the cross-beam profile measured at depth z_{max} should not differ by more than 3% for any pair of symmetric points with respect to the central ray.



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8.3 CLINICAL CONSIDERATIONS
8.3.1 Dose specification and reporting

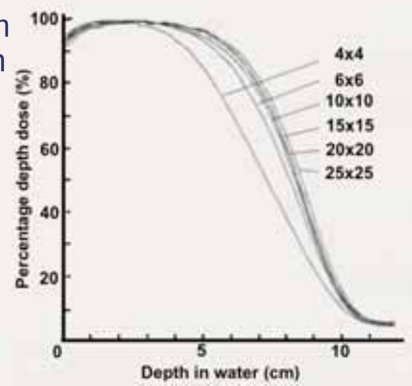
- Electron beam therapy is usually applied in treatment of superficial or subcutaneous disease.
- Electron beam treatment is usually delivered with a single direct electron field at a nominal SSD of 100 cm.
- The dose in electron beam therapy is usually prescribed at a depth that lies at, or beyond, the distal margin of the target.

8.3 CLINICAL CONSIDERATIONS
8.3.1 Dose specification and reporting

- To maximize healthy tissue sparing beyond the tumour and to provide relatively homogeneous target coverage treatments are usually prescribed at z_{max}, R_{90}, or R_{80}.
  - If the treatment dose is specified at R_{80} or R_{90}, the skin dose may exceed the prescription dose.
  - Since the maximum dose in the target may exceed the prescribed dose by up to 20%, the maximum dose should be reported for all electron beam treatments.

8.3 CLINICAL CONSIDERATIONS
8.3.2 Small field sizes

- The PDD curves for electron beams do not depend on field size, except for small fields where the side of the field is smaller than the practical range of the electron beam.
- When lateral scatter equilibrium is not reached at small electron fields, in comparison to a 10x10 cm^2 field:
  - Dose rate at z_{max} decreases.
  - Depth of maximum dose, z_{max}, moves closer to the surface.
  - PDD curve becomes less steep.



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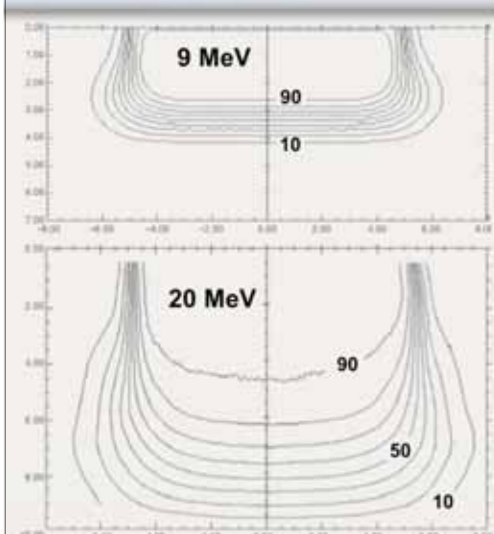
8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions



- Isodose curves are lines connecting points of equal dose in the irradiated medium.
- Isodose curves are usually drawn at regular intervals of absorbed dose and are expressed as a percentage of the dose at a reference point, which is usually taken as the z_{max} point on the beam central axis.

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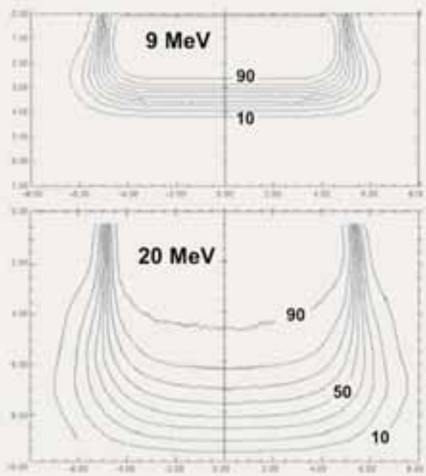
8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions



- As an electron beam penetrates a medium (absorber), the beam expands rapidly below the surface because of electron scattering on absorber atoms.
- The spread of the isodose curves varies depending on:
  - Isodose level
  - Energy of the beam
  - Field size
  - Beam collimation

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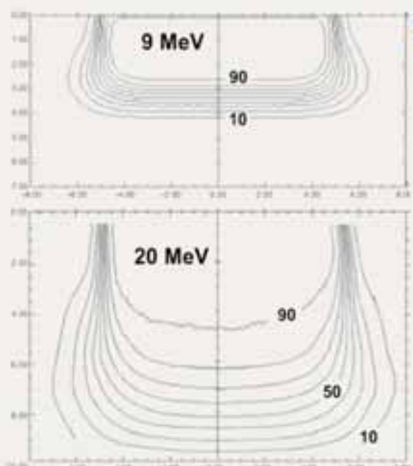
8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions



- A particular characteristic of electron beam isodose curves is the bulging out of the low value isodose curves (<20%) as a direct result of the increase in electron scattering angle with decreasing electron energy.

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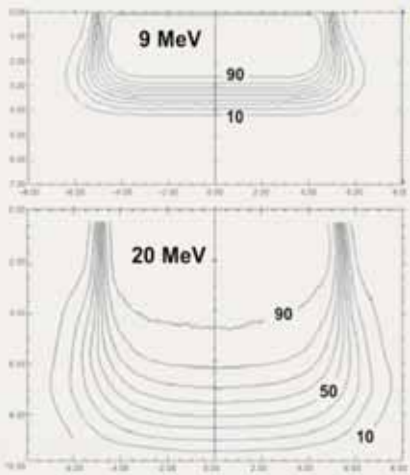
8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions

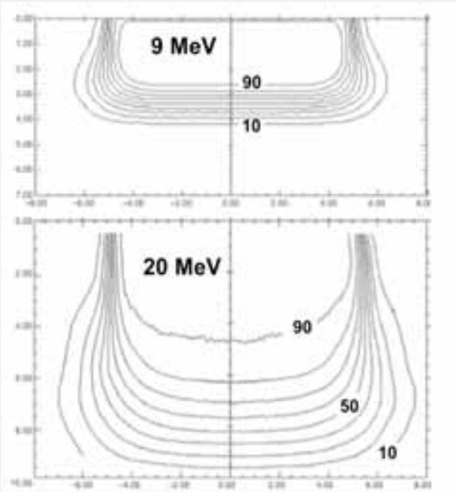


- At energies above 15 MeV electron beams exhibit a lateral constriction of the higher value isodose curves (>80%). The higher is the electron beam energy, the more pronounced is the effect.

8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions

- The term penumbra generally defines the region at the edge of the radiation beam over which the dose rate changes rapidly as a function of distance from the beam central axis.
- The physical penumbra of an electron beam may be defined as the distance between two specified isodose curves at a specified depth in phantom.

8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions



- In determination of the physical penumbra of an electron beam the ICRU recommends that:
  - The 80% and 20% isodose curves be used.
  - The specified depth of measurement be $R_{85}/2$, where R_{85} is the depth of the 85% dose level beyond z_{max} on the electron beam central ray.

8.3 CLINICAL CONSIDERATIONS
8.3.3 Isodose distributions

- In electron beam therapy, the air gap is defined as the separation between the patient and the end of the applicator cone. The standard air gap is 5 cm.
- With increasing air gap:
  - The low value isodose curves diverge.
  - The high value isodose curves converge toward the central axis of the beam.
  - The physical penumbra increases.

8.3 CLINICAL CONSIDERATIONS
8.3.4 Field shaping

- To achieve a more customized electron field shape, a lead or metal alloy cut-out may be constructed and placed on the applicator as close to the patient as possible.
- Field shapes may be determined from conventional or virtual simulation, but are most often prescribed clinically by a physician prior to the first treatment.
- As a rule of thumb, divide the practical range R_p by 10 to obtain the approximate thickness of lead required for shielding (<5%).

8.3 CLINICAL CONSIDERATIONS
8.3.4 Field shaping

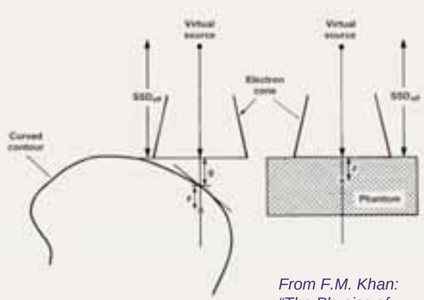
- For certain treatments, such as treatments of the lip, buccal mucosa, eyelids or ear lobes, it may be advantageous to use an internal shield to protect the normal structures beyond the target volume.
- Internal shields are usually coated with low atomic number materials to minimize the electron back-scattering into healthy tissue above the shield.

8.3 CLINICAL CONSIDERATIONS
8.3.4 Field shaping

- Extended SSDs have various effects on electron beam parameters and are generally not advisable.
- In comparison with treatment at nominal SSD of 100 cm at extended SSD:
  - Output is significantly lower
  - Beam penumbra is larger
  - PDD distribution changes minimally.
- An effective SSD based on the virtual source position is used when applying the inverse square law to correct the beam output at z_{max} for extended SSD.

8.3 CLINICAL CONSIDERATIONS
8.3.5 Irregular surface correction

- Uneven air gaps as a result of curved patient surfaces are often present in clinical use of electron beam therapy.
- Inverse square law corrections can be made to the dose distribution to account for the sloping surface.



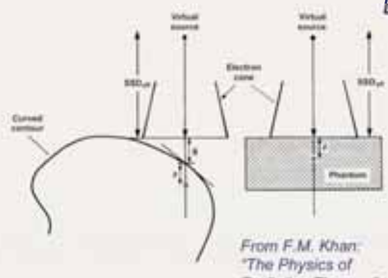
From F.M. Khan: "The Physics of Radiation Therapy"

$$D(\mathrm{SSD}_{\mathrm{eff}} + g, z) = D_{\mathrm{o}}(\mathrm{SSD}_{\mathrm{eff}}, z)\left\{\frac{\mathrm{SSD}_{\mathrm{eff}} + z}{\mathrm{SSD}_{\mathrm{eff}} + g + z}\right\}^2$$

g = air gap
z = depth below surface
$\mathrm{SSD}_{\mathrm{eff}}$ = distance between the virtual source and surface

8.3 CLINICAL CONSIDERATIONS
8.3.5 Irregular surface correction

- The inverse square correction alone does not account for changes in side scatter as a result of beam obliquity which:
  - Increases side scatter at the depth of maximum dose, z_{max}
  - Shifts z_{max} toward the surface
  - Decreases the therapeutic depths R_{90} and R_{80}.



From F.M. Khan: "The Physics of Radiation Therapy"

$$D(\mathrm{SSD}_{eff}+g,z) = D_o(\mathrm{SSD}_{eff},z)\left\{\frac{\mathrm{SSD}_{eff}+z}{\mathrm{SSD}_{eff}+g+z}\right\}^2 \mathrm{OF}(\theta,z)$$

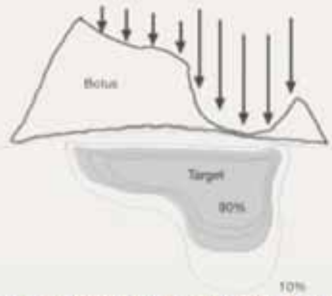
$\mathrm{OF}(z,\theta)$ = obliquity factor which accounts for the change in depth dose at a point in phantom at depth z for a given angle of obliquity θ but same SSD_{eff} as for $\theta = 0$

8.3 CLINICAL CONSIDERATIONS
8.3.6 Bolus

- Bolus made of tissue equivalent material, such as wax, is often used in electron beam therapy:
  - To increase the surface dose.
  - To shorten the range of a given electron beam in the patient.
  - To flatten out irregular surfaces.
  - To reduce the electron beam penetration in some parts of the treatment field.
- Although labour intensive, the use of bolus in electron beam therapy is very practical, since treatment planning software for electron beams is limited and empirical data are normally collected only for standard beam geometries.

8.3 CLINICAL CONSIDERATIONS
8.3.6 Bolus

- The use of computed tomography (CT) for treatment planning enables accurate determination of tumour shape and patient contour.
- If a wax bolus is constructed such that the total distance from the bolus surface to the required treatment depth is constant along the length of the tumour, then the shape of the resulting isodose curves will approximate the shape of the tumour as determined with CT scanning.



8.3 CLINICAL CONSIDERATIONS
8.3.7 Inhomogeneity corrections

- The presence of tissue inhomogeneities (also referred to as heterogeneities) such as lung or bone can greatly affect the dose distributions for electron beams .
- The dose inside an inhomogeneity is difficult to calculate or measure, but the effect of an inhomogeneity on the dose beyond the inhomogeneity is relatively simple to measure and quantify.

8.3 CLINICAL CONSIDERATIONS
8.3.7 Inhomogeneity corrections

- The simplest correction for for a tissue inhomogeneity involves the scaling of the inhomogeneity thickness by its electron density relative to that of water and the determination of the coefficient of equivalent thickness (CET).
- The electron density of an inhomogeneity is essentially equivalent to the mass density of the inhomogeneity.

8.3 CLINICAL CONSIDERATIONS
8.3.7 Inhomogeneity corrections

- CET is used to determine the effective depth in water equivalent tissue z_{eff} through the following expression:

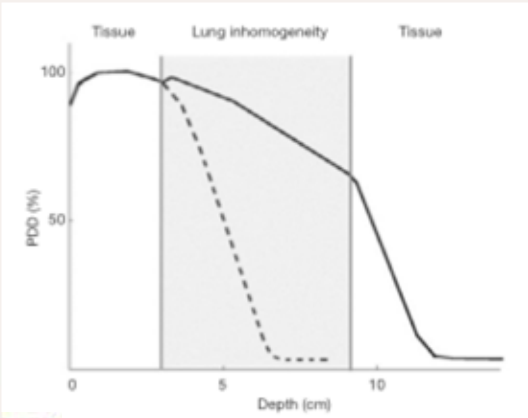
$$z_{eff} = z - t(1 - \text{CET})$$

z = actual depth of the point of interest in the patient
t = thickness of the inhomogeneity

- For example:
  - Lung has approximate density of 0.25 g/cm^3 and a CET of 0.25.
  - A thickness of 1 cm of lung is equivalent to 0.25 cm of tissue.
  - Solid bone has approximate density of 1.6 g/cm^3 and a CET of 1.6.
  - A thickness of 1 cm of bone is equivalent to 1.6 cm of tissue.

8.3 CLINICAL CONSIDERATIONS
8.3.7 Inhomogeneity corrections

- The effect of lung inhomogeneity on the PDD distribution of an electron beam (energy: 15 MeV, field: 10x10 cm^2).



Thickness t of lung inhomogeneity: 6 cm

Tissue equivalent thickness: z_{eff} = 1.5 cm

8.3 CLINICAL CONSIDERATIONS
8.3.7 Inhomogeneity corrections

- If an electron beam strikes the interface between two materials either tangentially or at a large oblique angle, the resulting scatter perturbation will affect the dose distribution at the interface.
- The lower density material will receive a higher dose, due to the increased scattering of electrons from the higher density side.

8.3 CLINICAL CONSIDERATIONS
8.3.7 Inhomogeneity corrections

- Edge effects need to be considered in the following situations:
  - Inside a patient, at the interfaces between internal structures of different density.
  - On the surface of a patient, in regions of sharp surface irregularity.
  - On the interface between lead shielding and the surface of the patient, if the shielding is placed superficially on the patient or if it is internal shielding.

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8.3 CLINICAL CONSIDERATIONS
8.3.8 Electron beam combinations

- Occasionally, the need arises to abut electron fields. When abutting two electron fields, it is important to take into consideration the dosimetric characteristics of electron beams at depth in the patient.
- The large penumbra and bulging isodose lines produce hot spots and cold spots inside the target volume.

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- In general, it is best to avoid using adjacent electron fields.
- If the use of abutting fields is absolutely necessary, the following conditions apply:
  - Contiguous electron beams should be parallel to one another in order to avoid significant overlapping of the high value isodose curves at depth in the patient.
  - Some basic film dosimetry should be carried out at the junction of the fields to ensure that no significant hot or cold spots in dose occur.

8.3 CLINICAL CONSIDERATIONS
8.3.8 Electron beam combinations

- Electron - photon field matching is easier than electron - electron field matching.
  - A distribution for photon fields is readily available from a treatment planning system (TPS) and the location of the electron beam treatment field as well as the associated hot and cold spots can be determined relative to the photon field treatment plan.
  - The matching of electron and photon fields on the skin will produce a hot spot on the photon side of the treatment.

8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- Electron arc therapy is a special radiotherapeutic treatment technique in which a rotational electron beam is used to treat superficial tumour volumes that follow curved surfaces.
- While its usefulness in treatment of certain large superficial tumours is well recognized, the technique is not widely used because it is relatively complicated and cumbersome, and its physical characteristics are poorly understood.

8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- The dose distribution in the target volume for electron arc therapy depends in a complicated fashion on:
  - Electron beam energy
  - Field width *w*
  - Depth of the isocentre d_i
  - Source-axis distance *f*
  - Patient curvature
  - Tertiary collimation
  - Field shape as defined by the secondary collimator

8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- Two approaches to electron arc therapy have been developed:
  - Electron pseudo-arc based on a series of overlapping stationary electron fields.
  - Continuous electron arc using a continuous rotating electron beam.
- The calculation of dose distributions in electron arc therapy is a complicated procedure that generally cannot be performed reliably with the algorithms used for standard electron beam treatment planning.

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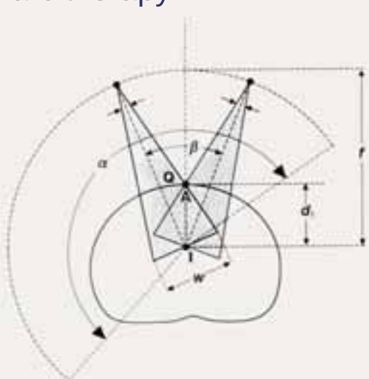
8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- The characteristic angle β concept represents a semi-empirical technique for treatment planning in electron arc therapy.



The characteristic angle β for an arbitrary point A on the patient surface is measured between the central axes of two rotational electron beams positioned in such a way that at point A the frontal edge of one beam crosses the trailing edge of the other beam.

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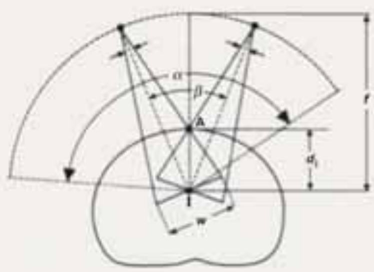
The characteristic angle β represents a continuous rotation in which a surface point A receives a contribution from all ray lines of the electron beam starting with the frontal edge and finishing with the trailing edge of the rotating electron beam.



- w is the nominal field size.
- f is the virtual source isocentre distance.
- d_i is the isocentre depth.

$$w = \frac{2d_i \sin\frac{\beta}{2}}{1 - \frac{d_i}{f}\cos\frac{\beta}{2}}$$

8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

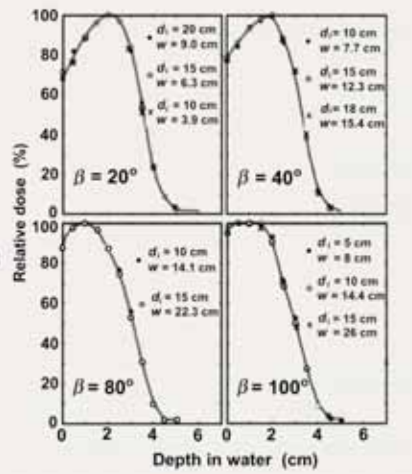
- The characteristic angle β is uniquely determined by three treatment parameters
  - Source-axis distance f
  - Depth of isocentre d_i
  - Field width w

$$w = \frac{2d_i \sin\frac{\beta}{2}}{1 - \frac{d_i}{f}\cos\frac{\beta}{2}}$$

- Electron beams with combinations of d_i and w that give the same characteristic angle β exhibit very similar radial percentage depth dose distributions even though they may differ considerably in individual d_i and w.

8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- The PDDs for rotational electron beams depend only on:
  - Electron beam energy
  - Characteristic angle β
  - When a certain PDD is required for patient treatment one may choose a β that will give the required beam characteristics.
  - Since d_i is fixed by the patient contour, the required β is obtained by choosing the appropriate *w*.



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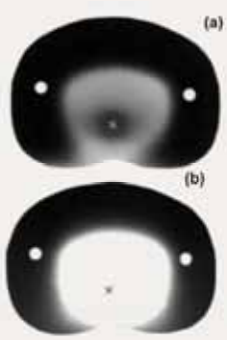
8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- Photon contamination of the electron beam is of concern in electron arc therapy, since the photon contribution from all beams is added at the isocentre and the isocentre may be at a critical structure.



Comparison between two dose distributions measured with film in a humanoid phantom:
(a) Small β of 10° (small field width) exhibiting a large photon contamination at the isocentre
(b) Large β of 100° exhibiting a relatively small photon contamination at the isocentre.

In electron arc therapy the bremsstrahlung dose at the isocentre is inversely proportional to the characteristic angle β.

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8.3 CLINICAL CONSIDERATIONS
8.3.9 Electron arc therapy

- The shape of secondary collimator defining the field width *w* in electron arc therapy is usually rectangular and the resulting treatment volume geometry is cylindrical, such as for example in the treatment of the chest wall.
- When sites that can only be approximated with spherical geometry, such as lesions of the scalp, are treated, a custom built secondary collimator defining a non-rectangular field of appropriate shape must be used to provide a homogeneous dose in the target volume.

8.3 CLINICAL CONSIDERATIONS
8.3.10 Electron therapy treatment planning

- The complexity of electron-tissue interactions makes treatment planning for electron beam therapy difficult and look up table type algorithms do not predict well the dose distribution for oblique incidence and tissue inhomogeneities.
- Early methods in electron beam treatment planning were empirical and based on water phantom measurements of PDDs and beam profiles for various field sizes, similarly to the Milan-Bentley method developed for use in photon beams.

8.3 CLINICAL CONSIDERATIONS
8.3.10 Electron therapy treatment planning

- The early methods in electron treatment planning accounted for tissue inhomogeneities by scaling the percentage depth doses using the CET approximation which provides useful parametrization of the electron depth dose curve but has nothing to do with the physics of electron transport.

- The Fermi-Eyges multiple scattering theory considers a broad electron beam as being made up of many individual pencil beams that spread out laterally in tissue following a Gaussian function.

8.3 CLINICAL CONSIDERATIONS
8.3.10 Electron therapy treatment planning

- The pencil beam algorithm can account for tissue inhomogeneities, patient curvature and irregular field shape.
  - Rudimentary pencil beam algorithms deal with lateral dispersion but ignore angular dispersion and backscat-tering from tissue interfaces.
  - Despite applying both the stopping powers and the scattering powers, the modern refined pencil beam, multiple scattering algorithms generally fail to provide accurate dose distributions for most general clinical conditions.

- The most accurate and reliable means to calculate electron beam dose distributions is through Monte Carlo techniques.
  - The main drawback of the current Monte Carlo approach to treatment planning is the relatively long computation time.
  - With increased computing speed and decreasing hard-ware cost, it is expected that Monte Carlo based electron dose calculation algorithms will soon become available for routine electron beam treatment planning.