Applications of Electron beam therapy:

The principal applications are (a) the treatment of skin and lip cancers, (b) chest wall irradiation for breast cancer, (c) administering boost dose to nodes, and (d) the treatment of head and neck cancers. Although many of these sites can be treated with superficial x-rays, brachytherapy, or tangential photon beams, the electron beam irradiation offers distinct advantages in terms of dose uniformity in the target volume and in minimizing dose to deeper tissues.

ELECTRON INTERACTIONS

As electrons travel through a medium, they interact with atoms by a variety of processes owing to Coulomb force interactions. The processes are (a) inelastic collisions with atomic electrons (ionization and excitation), (b) inelastic collisions with nuclei (bremsstrahlung), (c) elastic collisions with atomic electrons, and (d) elastic collisions with nuclei.

In inelastic collisions, some of the kinetic energy is lost as it is used in producing ionization or converted to other forms of energy such as photon energy and excitation energy. In elastic collisions, kinetic energy is not lost although it may be redistributed among the particles emerging from the collision. In low atomic number media such as water or tissues, electrons lose energy predominantly through ionizing events with atomic electrons. In higher atomic number materials, such as lead, bremsstrahlung production is more important. In the collision process with the atomic electrons, if the kinetic energy acquired by the stripped electron is large enough for it to cause further ionization, the electron is known as a secondary electron or a (delta)-ray. As a beam of electrons travels through a medium, the energy is continually degraded until the electrons reach thermal energies and are captured by the surrounding atoms.

Collisional Losses (Ionization and Excitation)

(a) The rate of energy loss depends on the electron density of the medium.

(b) (b) The rate of energy loss per gram per centimeter squared, which is called the mass stopping power, is greater for low atomic number (Z) material than for high Z materials. There are two reasons for this: First, high Z materials have fewer electrons per gram than low Z materials have and, second, high Z materials have more tightly bound electrons, which are not as available for this type of interaction.

(c) (c) The energy loss rate first decreases and then increases with increase in electron energy with a minimum occurring at about 1 MeV. Above 1 MeV, the variation with energy is very gradual.

(d) (d) The energy loss rate of electrons of energy 1 MeV and above in water is roughly **2 MeV/cm.**

Radiation Losses (Bremsstrahlung)

(e) The rate of energy loss per centimeter is approximately proportional to the electron energy and to the square of the atomic number (Z^2). Moreover, the probability of radiation loss relative to the collisional loss increases with the electron kinetic energy and with Z. That means that x-ray production is more efficient for higher energy electrons and higher atomic number absorbers.

Mean Energy:
The mean energy of the electron beam, Ä'$_0$, at the phantom surface is related to R_{50} (the depth at which the dose is 50% of the maximum dose) by the following relationship:
$E0=C_4.R_{50}$, where
C_4=2.33 MeV/cm in water & R_{50}=depth of 50% isodose

Determination of absorbed dose:
Calorimetry is the most basic method for the determination of absorbed dose, but because of technical difficulties, the use of calorimeters is not practical in a clinical setting. Ionization chambers and Fricke dosimeters are more commonly used. Film, thermoluminescent dosimeters (TLD), and solid state diodes are used to find the ratio of the dose at one point in a phantom to the dose at another point but not usually to measure the absolute absorbed dose at a point.

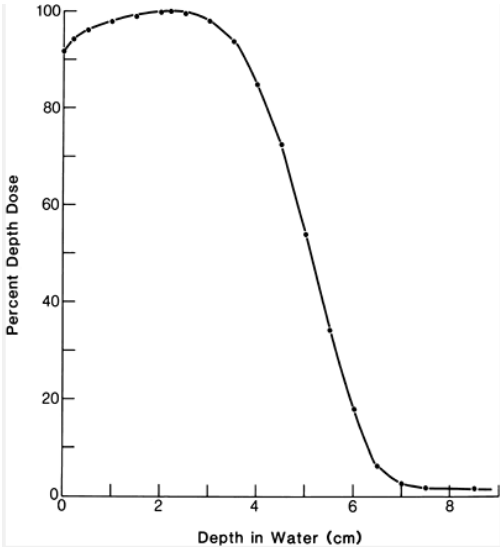
Output calibration:
Since the beam is calibrated to give 1 cGy/MU for the standard applicator at the depth of maximum dose on central axis (nominal SSD = 100 cm), the output factor for any applicator represents cGy/MU at d_{max}.

Phantoms:
Water is the standard phantom for the dosimetry of electron beams. However, it is not always possible or practical to perform dosimetry in a water phantom. For example, plastic phantoms are more suitable when using film or plane-parallel chambers. It also is difficult to make measurements near the surface of water, because of its surface tension and the uncertainty in positioning the detector near the surface.
For a phantom to be water equivalent for electron dosimetry it must have the same linear stopping power and the same linear angular scattering power. This is approximately achieved if the phantom has the same electron density (number of electrons per cubic centimeter) and the same effective atomic number as water. Of the commonly used materials for electron dosimetry, polystyrene and electron solid water (Radiation Measurements, Inc., Middleton, WI) come closest to being water equivalent.

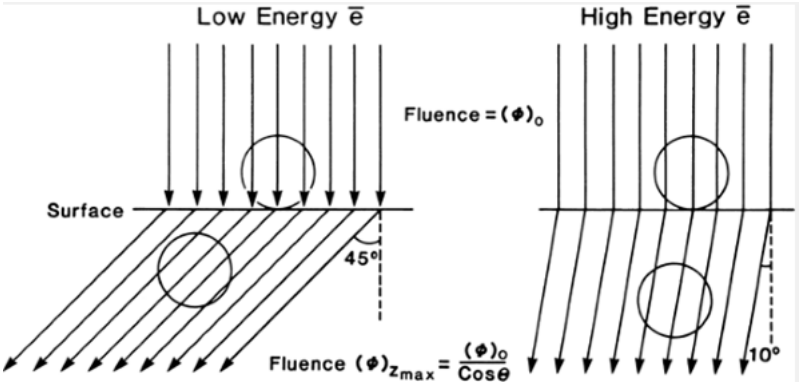
Central Axis Depth Dose Curve characteristic:

Beyond the maximum range of electrons, the dose is contributed only by the x-ray contamination of the beam, indicated by the tail of the depth dose curve.
For a broad beam, the depth in centimeters at which electrons deliver a dose to the 80% to 90% isodose level, is equal to approximately one-third to one-fourth of the electron energy in MeV.

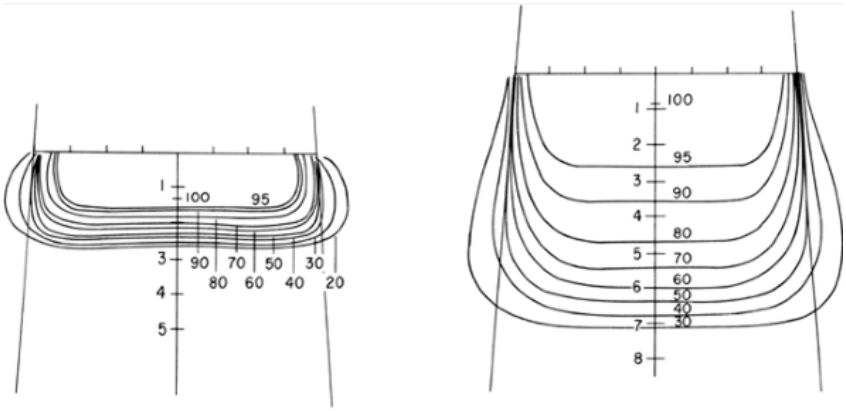
The most useful treatment depth, or therapeutic range, of electrons is given by the depth of the 90% depth dose. For modern accelerators with trimmer type applicators this depth is approximately given by E/3.2 cm, where E is the most probable energy in MeV of the electron beam at the surface. The depth of the 80% depth dose occurs approximately at E/2.8 cm. The depth of D_{max} does not follow a linear relationship with energy but it covers a broad region and its value may be approximated by 0.46 $E^{0.67.}$



The skin-sparing effect with the clinical electron beams is only modest or nonexistent. Unlike the photon beams, the percent surface dose for electrons increases with energy. This effect can be explained by the nature of the electron scatter. At the lower energies, the electrons are scattered more easily and through larger angles. This causes the dose to build up more rapidly and over a shorter distance. The ratio of surface dose to maximum dose is, therefore, less for the lower-energy electrons than for the higher-energy electrons.



Because of differences in beam generation, beam bending, and collimation, the depth dose distribution and the surface dose can be quite different for different machines.
Whereas for the low-energy beams all the isodose curves show some expansion, for the higher energies only the low isodose levels bulge out. The higher isodose levels tend to show lateral constriction, which becomes worse with decreasing field size.



7 Mev. Electron Beam
8cm. Circle. ∠1. F 5. 50 cm. TSD

18 Mev. Electron Beam
8cm. Circle, ∠ 5, F 7, 50cm. TSD.

Collimation:
All collimators provide a primary collimation close to the source that defines the maximum field size and a secondary collimation close to the patient to define the treatment field. The latter can be in the form of trimmer bars or a series of cones.
In the electron therapy mode, the x-ray collimator jaws are usually opened to a size larger than the cone or the applicator opening.
Such an arrangement minimizes the variation of collimator scatter, and therefore, the output variation with field size is kept reasonably small. If the collimator aperture (x-ray jaw setting) were allowed to change with the treatment field, the output would vary too widely with field size, especially for lower-energy beams.

Field size dependence:
As the field size is increased, the percent depth dose initially increases but becomes constant beyond a certain field size when the lateral scatter equilibrium is reached. Furthermore the depth d_{max} shifts toward the surface for the smaller fields.

The minimum field radius (in cm) for the establishment of lateral scatter equilibrium at all depths on central axis is given by $R_{eq}=0.88.\ \sqrt{E_{p,o,}}$where $E_{p,o,}$ is the most probable energy in MeV.

In clinical practice, the above relationship may be used to classify fields with radius < R_{eq} as small or narrow fields and radius >= R_{eq} as broad fields. The depth-dose distribution for small fields is field size dependent while for broad fields it is independent of field size.

Field size equivalence:
The term field equivalence means that for the same incident fluence and cross-sectional beam profile, the equivalent fields have the same depth-dose distribution along the central ray. Thus field equivalence here is defined in terms of percent depth doses and not the output factors, which depend on particular jaw setting for the given applicator or other collimation conditions. According to this definition, all broad fields are equivalent because their depth-dose distribution is the same irrespective of field size.
Field equivalence is therefore relevant only for small fields in which the lateral scatter equilibrium does not exist and consequently, the depth-dose distribution is field size dependent.

Virtual SSD:
Unlike an x-ray beam, an electron beam does not emanate from a physical source in the accelerator head. A pencil electron beam-after passing through the vacuum window of the accelerator, bending magnetic field, scattering foils, monitor chambers, and the intervening air column-is spread into a broad beam that appears to diverge from a point. This point is called the virtual source, which may be defined as an intersection point of the backprojections along the most probable directions of electron motion at the patient surface.
The use of virtual source-to-surface distance (SSD) does not give accurate inverse square law correction for output at extended SSDs under all clinical conditions.

An alternative method of correcting dose output for the air gap between the electron collimator and the patient is to determine effective SSD, which gives the correct inverse square law relationship for the change in output with distance. However, the effective SSD does change with energy and field size, especially for small field sizes and low energies.

X ray contamination:
The x-ray contamination dose at the end of the electron range can be determined from the tail of the depth-dose curve by reading off the dose value at the point where the tail becomes straight This dose in a patient is contributed by bremsstrahlung interactions of electrons with the collimation system (scattering foils, chambers, collimator jaws, etc.) and the body tissues.

For 5MeV, contributes 0.1% of Dmax dose, for 10MeV 0.5%, for 15 MeV 0.9%, for 20 MeV 1.4%.

The x-ray contamination is least in the scanning beam type of accelerator, because the scattering foils are not used. For regular treatment field sizes, the dose contributed by the x-ray contamination is not of much concern

Treatment planning:

Choice of energy: In most cases, when there is no danger of overdosing a critical structure beyond the target volume, the beam energy may be set so that the target volume lies entirely within the 90% isodose curve. However, in the treatment of the breast, the energy is often chosen so that the depth dose at the chest wall-lung interface is 80%.

Choice of field size: Examination of the electron isodose curves reveals that there is a significant tapering of the 80% isodose curve at energies above 7 MeV. The constriction of the useful treatment volume also depends on the field size and is worse for the smaller fields. Thus, with electrons, a larger field at the surface than one is usually accustomed to (in the case of photon beams) may be necessary to cover a target area adequately.

Use of bolus:
Bolus is often used in electron beam therapy to (a) flatten out an irregular surface, (b) reduce the penetration of the electrons in parts of the field, and (c) increase the surface dose. Ideally, the bolus material should be equivalent to tissue in stopping power and scattering power.

Adjacent fields:
Electron-Electron: When two adjacent electron fields are overlapping or abutting, there is a danger of delivering excessively high doses in the junction region. In a clinical situation, the decision as to whether the fields should be abutted or separated should be based on the uniformity of the combined dose distribution across the target volume. Because the tumors treated with electrons are mostly superficial, the electron fields are usually abutted on the surface. The hot spots can be accepted, depending on their magnitude, extent, and location.

Electron-Photon: When an electron field is abutted at the surface with a photon field, a hot spot develops on the side of the photon field and a cold spot develops on the side of the electron field. This is caused by outscattering of electrons from the electron field.

Field shaping:
For lower-energy electrons (<10 MeV), less than 5 mm thickness of lead is required for adequate shielding (e.g., <5% transmission).
Lead sheets of this thickness can be molded to conform more or less to the surface contour and, therefore, can be placed directly on the skin surface.
For higher-energy electrons, however, thicker lead is required and cannot be so easily contoured. Moreover, a heavy lead mask may cause discomfort to the patient. The alternative method is to support a lead cutout at the end of the treatment cone or the field trimmers.

Shields to be used in such a configuration can be designed from pure lead sheets or a low melting alloy such as Lipowitz metal.
Block thickness:Tthe minimum thickness of lead required for blocking in millimeters is given by the electron energy in MeV incident on lead divided by 2. Another millimeter of lead may be added as a safety margin.
The required thickness of Cerrobend is approximately 20% greater than that of pure lead.

Internal shielding:
In some situations, such as the treatment of lip, buccal mucosa, and eyelid lesions, internal shielding is useful to protect the normal structures beyond the target volume. Lead shielding
may be used to reduce the transmitted dose to an acceptable value. However, the electron backscatter from lead enhances the dose to the tissue near the shield. The enhancement in dose at the tissue-lead interface can be quite substantial, e.g., 30% to 70% in the range of 1 to 20 MeV, having a higher value for the lower-energy beams. To dissipate the effect of electron backscatter, a suitable thickness of low atomic number absorber such as bolus may be placed between the lead shield and the preceding tissue surface. eg an aluminum sheath around any lead used for internal shielding. Oral shielding has also been accomplished by special oral stents made of dental acrylic that encompasses the lead. Such a shield provides lead protection for the tongue and other structures as well as reduces the electron backscatter from lead reaching the buccal mucosa. Either 1 cm of bolus or 4 mm of aluminum may be used to absorb 90% of the backscattered electrons.
Electron arc therapy:
Electron beam arc technique gives excellent dose distribution for treating superficial tumors along curved surfaces. On the basis of isodose distribution, electron arc therapy is most suited for treating superficial volumes that follow curved surfaces such as chest wall, ribs, and entire limbs. Although all chest wall irradiations can be done with electron arcing, this technique is mostly useful in cases for which the tumor involves a large chest wall span and extends posteriorly beyond the midaxillary line. The conventional technique of using tangential photon beams in this case will irradiate too much of the underlying lung. The alternative approach of using multiple abutting electron fields is fraught with field junction problems, especially when angled beams are used. In short, it appears that for a certain class of cases, electron arc therapy has no reasonable alternative.
Treatment planning:
Beam energy: The central axis dose distribution is altered due to field motion. For a small scanning field width, the depth dose curve shifts slightly and the beam appears to penetrate somewhat farther than for a stationary beam. The surface dose is reduced and the bremsstrahlung dose at the isocenter is increased. This phenomenon is known as the “velocity effect”: a deeper point is exposed to the beam longer than a shallower point, resulting in apparent **enhancement** of beam penetration.

Scanning field width: Although any field width may be used to produce acceptable isodose distribution, smaller scanning fields (e.g., width of 5 cm or less) give lower dose rate and greater x-ray contamination. However, small field widths allow almost normal incidence of the beam on the surface, thus simplifying dosimetry. Another advantage of the smaller field width is that the dose per arc is less dependent on the total arc angle. For these reasons, a geometric field width of **4 to 8 cm at the isocenter** is recommended for most clinical situations.

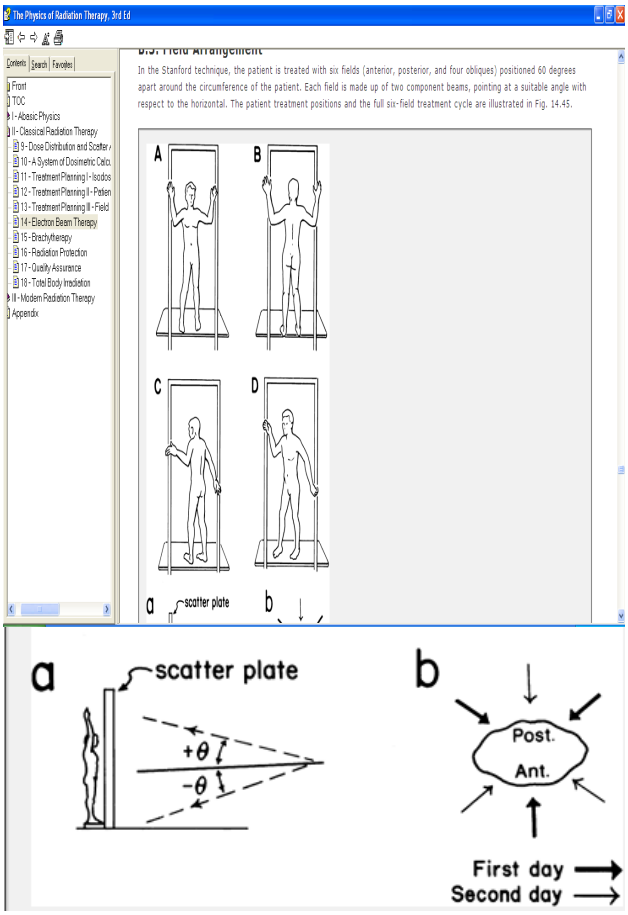
Location of isocentre: The isocenter should be placed at a point approximately equidistant from the surface contour for all beam angles. In addition, the depth of isocenter must be **greater than the maximum range** of electrons so that there is no accumulation of electron dose at the isocenter.

Field shaping: To sharpen the distribution, lead strips or cutouts should be used to define the arc limits as well as the field limits in the length direction.

Total Skin Electron Beam Therapy:
Basically, the methods fall into two general categories: (a) translational technique in which a horizontal patient is translated relative to a beam of electrons of sufficient width to cover the transverse dimensions of the patient and (b) large field technique in which a standing patient is treated with a combination of broad beams produced by electron scattering and large SSDs (2 to 6 m). The field is made uniform over the height of the patient by vertically combining multiple fields or vertical arcing. The patient is treated in a standing position with four or six fields directed from equally spaced angles for circumferential coverage of the body surface.

Reduction of X ray contamination:
In the Stanford technique, the electron beam, after emerging from the accelerator window, is scattered by a mirror (0.028-inch Al), an aluminum scatterer located externally at the front of the collimator (0.037-inch Al), and about 3 m of air before incidence on the patient.
The x-ray contamination incident on the patient is reduced by angling the beam 10 degrees to 15 degrees above and below the horizontal. Because the x-rays produced in the scatterers at the collimators are preferentially directed along the central axes, they largely miss the patient. In addition, this set-up provides a large electron field with sufficient dose uniformity in the vertical dimensions of the patient.

In the Stanford technique, the patient is treated with six fields (anterior, posterior, and four obliques) positioned 60 degrees apart around the circumference of the patient. Each field is made up of two component beams, pointing at a suitable angle with respect to the horizontal.



Beam energy:

When multiple large fields are directed at the patient from different angles, the composite distribution shows a net shift with apparent **decrease** in beam penetration. This shift of the relative depth doses closer to the surface is due to greater path lengths taken by the obliquely incident electrons in reaching a point.

Although a dose uniformity of ±10% can be achieved over most of the body surface using the six-field technique, areas adjacent to surface irregularities vary substantially due to local scattering. Areas such as inner thighs and axillae, which are obstructed by adjacent body structures, require supplementary irradiation.

Modified Stanford technique: The Stanford technique of six dual fields requires modifications of the accelerator such as removing the scattering foil and installing a scatterer at the front end of the collimator. These changes would require safety interlocks to prevent operation of the accelerator in this configuration for conventional electron beam treatments. Most institutions have adopted the Stanford technique in principle without making alterations in the accelerator hardware. Because the regular scattering foils and various interlocks are left in place, no special precautions are required in preparing the machine for total skin irradiation.

In some accelerators a high dose rate mode is installed to allow an output of more than 2,000 monitor units per minute. This significantly speeds up the treatments.

Because conventional electron cones are not used, the electron field is collimated by a special **wide-aperture insert** attached at the end of the collimator. It is preset via interlock to a wider jaw setting and a specific electron energy, selected for high dose rate mode of operation.
Some institutions use an **acrylic scatter plate** ([asymptotically equal to] 1 cm in thickness) in front of the patient to provide additional scatter to the electron beam.
To shorten the treatment time, the patient is treated with **three dual fields per day**, for example, day 1: one dual field from the anterior, two dual oblique fields from the posterior; day 2: one dual field posterior and two dual fields anterior oblique. A complete cycle of six dual fields is thus completed in **2 days**.
A source-to-patient distance of about **4 m** is sufficient for this technique.

Dual field angle:
A low-energy electron beam is considerably widened in size by scattering in air. For example, a 9-MeV electron beam, after transversing 4 m of air and an acrylic scatter plate, attains a Gaussian dose profile measuring a 90% to 90% isodose width of about 60 cm, which is usually sufficient to cover a patient's width. Along the height of the patient, two fields, one directed toward the head and the other toward the feet, are angled such that in the composite dose distribution ±10% dose uniformity can be obtained over a length of about 200 cm.

Dosimetry: *In vitro (Film):* The composite depth-dose distribution for the six dual fields may be determined by sandwiching a dosimetry film (in its paper jacket) in the cylindrical polystyrene phantom and cutting the excess film so that the edges conform to the circular surface of the phantom. A black tape is wrapped around the phantom over the film edges to make the film light-tight. The phantom, with the film parallel to the horizontal axis, is exposed to the six dual fields, duplicating actual treatment conditions. After appropriate processing, the film is scanned for optical density distribution, which is related to dose distribution by a reference sensitometric curve.

In vivo (TLD): Thermoluminescent dosimeters (TLD) are most often used for in vivo dosimetry. For these measurements, the TLD must be thin (<0.5 mm) to minimize the effect of dose gradient across the dosimeters.