CHAPTER V

CHARACTERISTICS OF ELECTRON BEAMS
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3.1 ELECTRON BEAM QUALITY AND CHARACTERISTICS

The increasing number of electron accelerators of various types with different beam qualities, require electron beam parameters to be defined for effective and unified approach of comparing these beams. Most often, the mere mention of a single parameter, namely energy of the beam, is not enough in the clinical situations. The energy of the beam is derived from the practical range of the electrons, which is generally related to the therapeutically useful interval of the absorbed dose distribution. It is also important from therapeutic point of view to define the treatment volume in a relevant and consistent way. Also proper diagnostics of the beam could be defined by a few independent parameters characterizing the beam quality. The absorbed dose distribution in a medium irradiated by a therapeutic electron beam of a given energy is dependent on the source to surface distance (SSD), to some extent on field size (especially for smaller field sizes), composition of the medium, angle of incidence, shape of the cone or applicator and the distance of applicator surface from the surface layer of the interacted medium.

5.1.1 DEPTH DOSE FOR ELECTRON BEAMS

The important interactions that are responsible for the depth dose distribution characteristics of electron beams are:
(a) inelastic collisions with orbital electrons, that is, ionization and excitation;
(b) inelastic collisions with nuclei, that is, bremsstrahlung;
and (c) elastic scattering, that is, Coulomb scattering with nuclei and orbital electrons.

Due to multiple scattering, the electrons follow tortuous path and almost continuously lose energy due to inelastic collisions as they travel in the medium. The electron energy spectrum changes continuously along with the fluence, with depth as the beam passes through any material. Andreo and Bielajew (1981) have shown that the shape of absorbed dose distribution is very similar to that of the fluence distribution of electrons. The initial build-up of fluence is due to the widening of the angular distribution of primary electrons with depth until the angular distribution is saturated, that is, full diffusion occurs. The rapid fall of the depth dose beyond $d_{\text{max}}$ is due to the increased loss of electrons as they are scattered into very oblique paths and run out of energy.

Various features which characterize the distribution of central axis depth absorbed dose for a broad therapeutic electron beam are:
(i) The surface dose;
(ii) The dose build up region; first the region of buildup of secondary electrons and to a lesser extent contamination with secondary photons and later the dose build up of primary electrons due to increased inclination of electron tracks;
(iii) Dose maximum where the electrons approach full diffusion;
(iv) The descending section which is the region of marked loss of primary electrons;
(v) The steep section of linear dose decrease due to energy and range straggling of the primary electrons;
(vi) The tail of the electrons having suffered few interactions in the first part of their tracks and
(vii) The photon background contributed by photons generated both in the phantom and in the parts of linac like vacuum window, scattering foils, ion chamber, x-ray collimators, cones, air etc.

5.1.2 SPECIFICATION OF ELECTRON BEAM ENERGY

Since electrons are charged particles, the electron energy spectrum changes continuously as the beam passes through any medium. The relatively narrow spectrum emerging from the vacuum exit window of the accelerator is characterized by $E_{p,a}$, the most probable energy in front of the vacuum exit window. The spectrum widens and decreases in energy as the beam passes through scattering foils, monitor chamber and air along its path. At the phantom or patient surface, the spectrum is characterized by two parameters, the most probable energy at the surface, $E_{p,o}$, which corresponds to the peak of the energy spectrum and the mean energy at the surface, $E_o$. Because the interactions involving loss in electron energy are such as to skew the spectra towards lower energies, $E_o$ is always less than $E_{p,o}$, the difference increases with the energy spread of the spectrum. Thereafter, as
the beam penetrates further, the energy continues to decrease and the mean energy at depth \( Z \), \( E_Z \), falls, until the beam is completely stopped. The specification of energy is obtained from empirical relationships between electron energy and the measured values of certain characteristics of the depth attenuation curve in water.

5.1.3 MOST PROBABLE ENERGY AT THE SURFACE (\( E_{p,0} \))

Most probable energy at the surface, \( E_{p,0} \) is a useful parameter to characterize depth dose distribution as it is well related to the practical range, \( R_p \), and it is the rounded value of this energy parameter which is set at the machine. The most probable energy at the surface \( E_{p,0} \) in MeV is related to practical range \( R_p \) in cm in water by

\[
E_{p,0} = C_1 + C_2 R_p + C_3 R_p^2
\]

where \( C_1 \), \( C_2 \) and \( C_3 \) are constants.

NACP (1981) and ICRU (1984) found that with \( C_1 = 0.22 \) MeV, \( C_2 = 1.98 \) MeV cm\(^{-1} \) and \( C_3 = 0.0025 \) MeV cm\(^{-2} \) reproduces measured data within an accuracy of 2% from few MeV to 50 MeV. Task group 25 (AAPM) recommends that these values are valid for broad beams which are greater than 12x12 cm\(^2 \) for energies up to 20 MeV and greater than 20x20 cm\(^2 \) for higher energies for SSD's greater than 100 cm. Recent Monte carlo simulations using EGS4 code have generated similar values (NACP 1980). To be in strict accordance with this formalism, \( R_p \) should be determined from measured depth.
dose data corrected for divergence. However, if one uses depth-
ionization curves for SSD greater than 100 cm and uncorrected for
divergence, difference in $R_p$ is not clinically significant for
the calculation of $E_p,0$. Task group 25 takes the position that
$R_p$ may be determined from depth dose or depth ionization curves
with or without divergence corrections for clinical applications.
A broad beam narrow detector geometry is recommended by the Task
Group 25 (1991). In order to correct for beam divergence, the
measured values of dose or ionization at depth should be multi-
plied by the factor \((SSD_{\text{eff}} + d)^2/SSD_{\text{eff}}^2\) before determination of
range. \(SSD_{\text{eff}}\) is the effective source to phantom surface dis-
tance corrected for virtual source positions (NACP 1980).

Energy spectrum of the high energy electron beam, unlike
that of x-ray beam, decreases in energy rapidly with depth in the
absorbing medium, with a consequent change in stopping power with
depth. The phenomenon originally postulated by Fermi called
polarization density effect is shown to be important for high
energy electrons, as the consequence that the dose conversion
factor for a gaseous ionization chamber in water or other medium,
varies with depth. Stopping power tables in literature include
this effect (ATP for Varian user's).

i.1.4 MEAN ENERGY AT THE SURFACE ($\bar{E}_0$)

The most significant energy parameter for dosimetry is $E_0$, which
is the mean energy at the surface. ICRU 21 (1972) recom-
ends the following relationship for its determination
\[ E_0 = C_4 \times R_{50} \] where \( R_{50} \) is the depth of 50\% dose or ionization and \( C_4 \) is 2.33 cm\(^{-1}\) for water in the energy region of 5 to 35 MeV. The factor 2.33 MeV cm\(^{-1}\) was originally obtained from Monte Carlo calculations by Rogers and Bielajew (1990) from depth dose curves for mono-energetic broad parallel beams, therefore, the relationship should strictly be,

\[ E_0 = 2.33 \times R_{50,d} \]

The super-script implying that the curve is measured for a constant source-to-chamber distance (SCD). It is often more convenient to use measured values of \( R_{50} \) taken from depth dose curves at a fixed SSD \((f)\) to give \( R_{50,f} \) or from depth ionization curves \((R_{50,f,i})\) usually \( f=100 \). There are differences between the \( R_{50} \) values obtained from these approaches, which increases with energy, and strictly correction should be applied to the equation if either of the other two \( R_{50} \) values are used. Curves are presented in NACP protocol (1980) showing the relationship between \( E_0 \) and \( R_{50} \) obtained from depth dose or depth ionization curves at 100 cm SSD. This protocol also indicated that the same expression can be used with \( R_{50,f} \) for energies from 5 MeV upto about 20 MeV. Table (iv) of IAEA protocol, TRS 277 (1987) presents tabulated values of these data linking \( E_0 \) to both \( R_{50,f} \) and \( R_{50,i} \).

The value of \( C_4 \) linking \( E_0 \) and \( R_{50,i} \) is 2.38 MeV cm\(^{-1}\) HPA (1985) and IPSM (1992) protocols and AAPM recent report (1983) recommend a value of \( C_4 \) to be equal to 2.4 MeV cm\(^{-1}\) for the range of energies used clinically. HPA protocols use 2.4 MeV cm\(^{-1}\) recommending either measurements fixed SCD or appropriate correc-
tions applied to fixed SSD measurements. It may be noted that AAPM protocol uses equation $E_0 = 2.33 \times R^f_{50,1}$ applied to $R^f_{50,1}$. For any of these $R_{50}$ determination, sufficiently large field sizes should be used, such that the values are independent of field size. $E_0$ thus obtained must be applied to all field sizes with that beam, including smaller fields where the depth dose curves are altered due to reduced scatter from smaller irradiated area. Field sizes of at least $12 \times 12$ cm$^2$ to be used up to 15 MeV and larger for the higher energies. All depths must be corrected to take into account the effective point of measurement of chamber.

5.1.5 MEAN ENERGY AT DEPTH $Z$ ($\bar{E}_z$)

For absorbed dose determination with an ion chamber, it is necessary to know the mean energy at the location of the detector in order to determine the replacement correction. A widely used relationship, based on mono-energetic electrons having a surface energy of $E_0$ is

$$\bar{E}_z = \bar{E}_0 \left(1 - Z/R_p\right)$$

Where the depth of interest $Z$ and $R_p$ are measured in the same material. This approximate relationship is close to the true values only for lower energy electron beams, or at depths close to the surface and close to $R_p$ for higher energies. Monte carlo calculations have been carried out as the realistic variation of $E_z$ with depth, which is summarised in IAEA protocol (1987).
The energy parameter $E_{p,0}$, $E_0$, and $E_z$ as well as $R_5$ and $R_{50}$ are of direct interest in treatment planning and dosimetry. A depth dose or depth ionization measurement for every electron beam energy is a must to get all these parameters.

1.1.6 ELECTRON BEAM SYMMETRY AND FLATNESS

Symmetry and flatness of an electron beam depends on the design and adjustment of the flattening system and the applicators used for beam collimation. Uniformity of intensity across the electron beam is necessary for therapeutic use of electron beams. After the electron beam passes through the accelerator exit window, it comes across the scattering and flattening foils, beam monitoring chambers, x-ray beam collimator jaws, electron applicator, air etc. Most often scattering from the cones and x-ray collimator jaws is used for beam flatness. Since the scattered electrons are of lower energy, the flatness of the beam may change significantly with depth (NACP 1980). Electron beam uniformity can be defined by setting specific recommendations on field symmetry and flatness for a specific area of the beam at a depth, which is generally the depth of maximum dose. It is recommended AAPM TG 25, 1991) that the uniformity also be evaluated near the surface and at the therapeutic range. The measurements should be made on each of the principal and diagonal axis of the beam for the largest field for each applicator. Flatness specifications
should be checked at several collimator angles and at several gantry angles.

Since the depth of dose maximum varies significantly for different machines, Task Group 25 (1991) recommends that the reference plane be at the depth of 95% isodose beyond $d_{\text{max}}$. Variations of dose normalized to the central-axis value should not exceed $\pm 5\%$ (optimally $\pm 3\%$) over an area confined within lines 2 cm inside the geometric edge of fields equal to or larger than 10x10 cm$^2$. Regarding beam symmetry, the cross beam dose profile in the plane of reference should not differ more than 2% at any pair of points situated symmetrically, with respect to the central ray.

5.1.7 ELECTRON BEAM CHARACTERISTIC PARAMETERS

The characteristic parameters of the electron beams as defined mainly in AAPM Task Group 25's report, (1991) and ICRU report #35 (1984) are listed below:

5.1.7.1 RANGE PARAMETERS

'100'-depth of maximum dose.

'90'-depth of 90% dose, beyond the dose maximum, also defined as the therapeutic range in the AAPM Task Group 25's report.

'85'-depth of 85% dose, beyond the dose maximum, also the therapeutic range as defined in the ICRU Report #35.
$R_{90-10}$ is an indicator of dose fall-off beyond the 90% dose level.

$R_{85-10}$ is the indicator of dose fall-off beyond 85% dose level.

$R_{50,R10}$ are the range parameters for the depth of 50% and 10% dose levels.

$r_p$ is the depth at which the tangent at the 50% dose point intercepts the extrapolated photon dose level and is known as the practical range.

$r_q$ is the depth at which the tangent at the 50% dose point intercepts the horizontal line through the point of dose maximum.

### 5.1.7.2 CENTRAL AXIS DOSE

The following parameters were first defined by Brahme and Svensson (1976) and later adopted by ICRU (1984):

G- normalized dose gradient which is the measure of the steepness of the absorbed dose distribution $= \frac{R_p}{(R_p-R_q)}$

$D_p/D_m$ is the ratio of surface to the maximum dose, surface dose is measured at a depth of 0.5 mm.

$D_x/D_m$ is the ratio of the extrapolated photon dose to the dose maximum, the photon background $D_x$ is defined as an extrapolation of the tail of the absorbed dose distribution back to the practical range.

All the parameters mentioned above are also depicted in the Fig.5.1.
Fig. 5.1 Depth dose curve and the beam parameters for an electron beam.
5.1.7.3 DOSE IN PLANE PERPENDICULAR TO THE CENTRAL AXIS

The plane selected here is at a depth of half the therapeutic range i.e. $\%R_{85}$ and the dose in this plane is normalized to 100% at the centre. A typical distribution is shown in Fig.5.2. Dosimetric parameters relevant to this are addressed below:

$U_{90/50}$ is the ratio of the area inside the 90% isodose line to that inside the 50% isodose line and is called as uniformity index.

$P_{80/20}$ is the average distance separating the 80% and 20% isodose lines and is a measure of the radiation penumbra.

5.1.7.4 RELATIONSHIP BETWEEN ENERGY AND RANGE PARAMETERS

$E$-nominal beam energy

$E_0$-mean energy at the surface = 2.33 $R_p$.

$E_{p,0}$-most probable beam energy at the surface

$E_{p,0} = 0.22 + 1.98 R_p + 0.0025 R_p^2$.

$E_0$-single energy value of the electron beam at the surface assuming that the total energy loss in layers traversed by the beam is small (1984) $E_0 = (R_p + 0.376)/0.521$
Fig. 5.2 Isodose distribution for a 9 MeV electron beam in a plane perpendicular to the central axis at the depth of $D_{\text{max}}$. 
5.1.7.5 DETERMINATION OF ABSORBED DOSE

The absorbed dose to air chamber factor is defined as (TRS 277)

\[
N_{d,u} = \frac{D_{\text{air,u}}}{M_u}
\]

\[N_{d,u} = N_k \times (1-g) \times k_{\text{att}} \times k_{\text{m}} \times k_{\text{air,c}}\]

Where, \( N_k \) is the kair calibration factor of the ionization chamber
\( k_{\text{att}} \) is the fraction of energy of secondary charged particle that is lost to bremsstrahlung.
\( k_{\text{m}} \) is the SWlair-stopping power ratio water to air.

\( D_{\text{w}}(p_{\text{eff}}) = D_{\text{air,u}} \times (S_{\text{w,air}}) \times P_u \)

\( D_{\text{w}} \)-absorbed to water

\( k_{\text{air}} \)-air kerma

\( D_{\text{air}} \)-mean absorbed dose inside the air cavity.

\( N_k \)- air kerma calibration factor of the ionization chamber

Start of Subsection 5.1.7.6

Perturbation Correction Factor (Pu) corrects for the different properties in electron production and scattering in the chamber wall and corresponding volume of water and it accounts for the difference in electron scattering in the air cavity and in the water which is "replaced" by the air cavity.
Factor $k_m$ takes into account the lack of air equivalent of the ionization chamber material at the calibration in Co-60 gamma ray beam.

Factor $k_{att}$ takes into account the attenuation and scattering of the photons in the ionization chamber material, including the build-up cap. The influence of the photons scattered and absorbed in the chamber is thus included in this factor.

Experimental values of the product $k_m k_{att}$ are available in the protocol tables. Corrections for the non-air equivalence of the central electrode have been disregarded.

5.1.7.6 EFFECTIVE POINT OF MEASUREMENT ($P_{eff}$)

The photon or electron fluence in the uniform water phantom is perturbed in the volume which is occupied by the ionization chamber when measurements are performed. This effect is corrected by the use of an effective point of measurement that takes into account the spatial extent of the air by locating the point of interest $P_{eff}$ towards the source of electrons/photon beams above the chamber centre to correct for the gradient of fluence within the chamber cavity. The effective points of measurement $P_{eff}$ for the high energy x-ray is 0.75$r$ earlier to the cylindrical chamber center, and for the high energy electrons it is 0.5$r$ in front of the cylindrical chamber centre, where $r$ is the internal radius of the chamber. For plane parallel chamber, the front surface of the air cavity is the effective point of measurement.
5.2 ELECTRON BEAM OUTPUT FACTORS AS A FUNCTION OF RADIOThERAPY PORTAL AREAS

5.2.1 SCIENTIFIC BACKGROUND:

High energy electron beams are preferred for radical Radiotherapy because the dose is relatively uniform from the surface to a given depth, the depth of penetration can be controlled by altering the incident electron energy or by using tissue compensators and the mass stopping power of electrons does not vary significantly for normal tissues. The electron beams in the range of 6-20 MeV has a region of more or less uniform dose followed by a rapid fall off of dose, which made it possible to use single field irradiations without delivering significant doses to the deep lying structures. Due to the rapid fall off dose beyond 90% dose level, the electron energy selection to treat a tumour should be arrived at carefully. The dose level within the first few millimeters of the tissue approaches 90% of the dose at $d_{\text{max}}$, which indicates that there is only modest skin sparing effects with clinical electron beams. Unlike photon beams, the surface dose for electron beams increases with energy due to the difference in scatter angles.

The output factor of an electron beam is defined as the ratio of maximum dose on the central axis of the field of interest to that of a reference field. The output factor variation for the clinical electron beam with field size is greater than that
exhibited by high energy photon beams. For the photon beams the change in output factor is due to the change of secondary or scattered photon components. As this changes are much smaller than the primary photon components, a less field size dependency is noticed. Since there is no primary component to the dose in the electron beam and all the dose at \( d_{\text{max}} \) is due to the multiple scattered electrons, there is a strong field size dependence on the output factor. The variation in output is primarily due to electrons scattering between the scattering point and the patient. The amount of scattering is field size and shape dependent and therefore be strongly influenced by the nature of the collimating system (Almond 1976, Biggs et al 1979, Choi et al 1979, De almeida 1973, Goede et al 1977, Khan et al 1987). Reducing the field size significantly alters the maximum dose value and the shape of the depth dose curve (Bruinvis et al 1983, De almeida 1973, McGinley et al 1979, Sharma et al 1984, Svensson 1971). The alteration of the dose becomes stronger when the beam contains electrons scattered by collimating devices, particularly by applicator walls and field defining frames (ICRU 1984, Lax 1980, Hills et al 1972, 1985, Meyer et al 1974). Because irregularly shaped electron fields are used to treat many clinical disorders, there is a need to quantify the variation of output for irregular fields defined in a standard applicator supplied by the manufacturer. In this study, we have attempted to quantify the effect of different cones on the output for various electron energies and the output variations for different field sizes defined in the standard applicator.
5.2.2 MATERIALS AND METHODS

The electron applicators provided by the manufacturer along with Clinac 1800 can produce field sizes varying from 4x4 to 25x25 cm at isocenter. The electron beams are collimated by the photon jaws and the set of electron applicators after they pass through the scattering foil. The photon collimator openings vary with applicator size and beam energy. (Refer Table 5.1).

<table>
<thead>
<tr>
<th>Applicator size (cm²)</th>
<th>Electron energy (MeV)</th>
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<tbody>
<tr>
<td></td>
<td>9</td>
</tr>
<tr>
<td>4x4</td>
<td>20x20</td>
</tr>
<tr>
<td>6x6</td>
<td>20x20</td>
</tr>
<tr>
<td>10x10</td>
<td>20x20</td>
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</tr>
<tr>
<td>25x25</td>
<td>30x30</td>
</tr>
</tbody>
</table>

Square fields of varying dimensions were obtained using the machined cerrobend shielding inserts in a 20x20 cm standard applicator.
Ionometric measurements were carried out with 0.6cc Farmer type ion chamber (PR-06G, Capintec Inc., USA) in a polystyrene phantom at the depth of maximum ionization. Four cerrobond inserts which can produce square fields of 4x4, 6x6, 10x10 and 15x15 cm were used for the estimation of relative output factor. The output factor was calculated by normalizing each measured output to the output at d\textsubscript{max} for a 10x10 cm field.

The outputs were measured for all the open cones, without field shaping at each energy. Measurements were also made with the cerrobond inserts at all energies. The output factors were plotted for the square fields for different energy electron beams normalized to 10x10 cm cone. All the measurements were made with the recommended X-ray collimator setting (Table 5.1) for the particular cone regardless of the size of the insert.

5.2.3 RESULTS

The curves in Fig.5.3 shows the relative change in output factor as function of side of the square field. The variation of output factor does not show any regular pattern for different cones. However, there is an increase in output of 3% to 5% from 4x4 cm to 10x10 cm square fields. The output further drops up to 3% in low energy electrons (up to 9MeV) compared to 1% in high energy electrons (12MeV-20MeV). The 20 MeV output factors decreases about 10% for a smaller field size (4x4 cm) from the 10x10 cm output factor. When the cerrobond inserts were placed in a 20x20 cm
Fig. 5.3 Variation of electron output factor with field sizes defined by standard cones.

Fig. 5.4 Variation of electron output factor with field sizes defined by cerrobend cut-outs in a 20x20 cm cone.
Fig. 5.5 Variation of output factor with field opening normalized to an output of 20x20 cm standard cone.

Fig. 5.6 Variation of output factor with the percentage of field shielded in a standard 20x20 cm cone.
standard cone (cone 40), the output factor (Fig.5.4) for the fields so obtained does not follow the open cone output factor curves. Instead, the shaped fields have their own output factors. It was observed that, when an insert is placed into a cone where the insert has the same size as the open cone, a change is observed in output factor. The magnitude of this effect is more pronounced at the higher energies and tends in most cases to decrease with increasing field size for each energy. The output factor for the shaped fields almost remains constant within 4% except for high energy electrons (16 and 20 MeV), as long as no more than three fourth of the field is blocked (Fig.5.5). It is also observed that the output factor form smooth and continuous curves as a function of field size (except 20 MeV electrons) which is shown in Fig.5.6.
5.3 EFFECT OF SHIELDING ON VIRTUAL SOURCE POSITION FOR ELECTRON BEAMS

5.3.1 SCIENTIFIC BACKGROUND:

The pencil electron beam from the linear accelerator head after passing through the exit window, bending magnetic field, scattering foils, monitor chambers, cone and the intervening air column, gets spreaded into a broad beam which appearing to diverge from a point, which is called the virtual source (Fig.5.7). Unlike the X-ray beam which originates from a distinct focus, the X-ray target, electron beam appears to come out of a source which is at distance not coinciding with the scattering foil or the accelerator exit window. The International Commission on Radiation Units and Measurements (ICRU 1984) defines the virtual source as a source which when placed in vacuum at some distance from the phantom surface produces exactly the same electron fluence as the real beam (ICRU 1984). The knowledge of virtual source position for an electron beam is necessary for an absolute dose estimation and treatment planning computations. Whenever a surface irregularity or curvature is encountered in radiotherapy portal, the dose at a point within the patient has to be quantified applying the inverse square law correction, for which the virtual source position need to be known accurately.

The virtual point source position can be estimated using Full-Width at Half-Maximum (FWHM) method, (Hogstrom 1985, Khan et al 1978, Meyer et al 1984) or by Multi Pin-hole Camera (MPC).
Fig. 5.7 Virtual source position of an electron beams defined by a fixed electron applicator.
method (Schroder Babo 1983). These methods give consistent results only for large field sizes and for energies greater than 15 MeV. The Inverse-Square Law (ISL) method (Almond 1976), the Inverse-Slope (IS) method (Khan et al 1978) and the Power Law (PL) method (Sweeney et al 1981) are usually preferred, as these methods nearly simulate the clinical conditions.

The virtual point source position of the electron beams must be determined for each individual accelerator as the geometrical and the scattering conditions vary significantly. Many studies have been carried out (Faermann et al 1983, Ghazi and Lingman 1991, Jamshidi et al 1986, Khan 1986, Klevenhagen 1985, Sharma et al 1992, Yudelev et al 1982) to find out the virtual source position for different accelerators for the standard cones supplied by the manufacturers. The Inverse Square Law (ISL) or extrapolation method is relatively insensitive to changes in beam collimation (Khan 1986). In the present study the virtual point source positions were determined for various electron energies and field sizes defined by the standard electron cone supplied by the manufacturer, using ISL method.

1.3.2 MATERIALS & METHODS:

The electron applicators supplied along with the machine have field sizes varying from 4x4 to 25x25 cm² at the isocentre. After passing through the scattering foil the electron beams are collimated by the photon jaws and the set of electron applicators. The photon collimator openings vary with applicator size
beam energy which is shown in Table-5.1. Different square fields were obtained using machined cerrobond shielding inserts. A 20x20 cm² standard applicator were machined.

Ionometric measurements were carried out with a 0.6cc Farmer type ion chamber (PR-06G Capintec Inc., USA) at the depth of maximum ionization in a polystyrene phantom. Square fields of 4x4, 6x6, 10x10 and 15x15 cm² were made available using four cerrobond inserts for the estimation of virtual source positions. Nominal source to surface distances (SSD) were plotted against inverse of the square of the charge. The extrapolation of the curve back to the abscissa yields the virtual source position.

5.3.3 RESULTS:

The effect of field size on the virtual source position is indicated in Fig.5.8, Fig.5.9 and Fig.5.10 for different electron energies (6, 12, and 20 MeV). The measured virtual source position with respect to the isocentre for the standard cones supplied by the manufacturers are shown in Table-5.2. The virtual SSD's for the standard 20x20 cm² cone and for the fields defined by placing the cerrobond shields are summarised in Table-5.3.

The virtual source positions were observed to change with the beam energies and the applicator sizes which is indicated in Fig.5.11. The variation of virtual source distances with the field opening in the standard applicator (20x20 cm²) is shown in Fig.5.12.
Fig. 5.8 Graph of SSD versus inverse of square root of charge for 4x4 cm field opening in a standard 20x20 cm cone.

Fig. 5.9 Graph of SSD versus inverse of square root of charge for 6, 9, and 12 MeV electrons for standard 20x20 cm cone.
Fig. 5.10 Graph of SSD versus inverse of square root of charge for 15x15 cm field opening in a standard 20x20 cm cone.
Fig. 5.11 Variation of virtual source position as a function of field size for different electron energies.

Fig. 5.12 Variation of virtual source position as a function of field opening.
### TABLE 5.2

Distances from the virtual sources to the isocentre for standard cones of various energies

<table>
<thead>
<tr>
<th>Applicator size (cm²)</th>
<th>Nominal Electron Energy (MeV)</th>
<th>6 Mev</th>
<th>9 MeV</th>
<th>12 MeV</th>
<th>16 MeV</th>
<th>20 MeV</th>
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<tr>
<td>4x4</td>
<td>63.5</td>
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<td>96.1</td>
<td>97.7</td>
<td>98.0</td>
<td></td>
</tr>
</tbody>
</table>

### TABLE 5.3

Distances from the virtual sources to the isocentre for different field sizes in a standard 20x20 cm² cone for various energies

<table>
<thead>
<tr>
<th>Applicator size (cm²)</th>
<th>Virtual source positions for (cm)</th>
<th>6 Mev</th>
<th>9 MeV</th>
<th>12 MeV</th>
<th>16 MeV</th>
<th>20 MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>4x4</td>
<td>47.2</td>
<td>60.4</td>
<td>71.7</td>
<td>75.5</td>
<td>80.0</td>
<td>82.7</td>
</tr>
<tr>
<td>6x6</td>
<td>62.6</td>
<td>65.4</td>
<td>74.7</td>
<td>76.2</td>
<td>82.3</td>
<td>83.9</td>
</tr>
<tr>
<td>10x10</td>
<td>70.8</td>
<td>74.0</td>
<td>83.9</td>
<td>82.3</td>
<td>91.6</td>
<td>97.6</td>
</tr>
<tr>
<td>20x20</td>
<td>83.5</td>
<td>86.4</td>
<td>96.1</td>
<td>97.7</td>
<td>98.0</td>
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</tr>
</tbody>
</table>

(Std.App)
5.4 DOSIMETRIC ACCURACY AT LOW MONITOR UNIT SETTING AT DIFFERENT DOSE RATES

4.1 SCIENTIFIC BACKGROUND:

Dose delivery of ±5% to the target volume for better tumour control (AAPM 1983, WHO 1988) can be achieved only when an uncertainty of ±2% in radiation dosimetry is maintained. Electron beam still remains the best choice of single field treatment for the treatment of tumours near the body surface. For Segmental and hyperfractionated radiotherapy, electron beams with low monitor unit (LMU) setting is necessary (Jenson 1990). Also a few dosimetric studies relating to TLD phosphors for radiation protection, film dosimetry at low dose level and research applications related to environmental dosimetry, nuclear medicine dosimetry and beta dosimetry requires an electron beam with low MU setting. Hence better understanding of accuracy in dose delivery for electron beams at LMU is necessary. Das et al (1991), Sharma et al (1994), Rajapakshe and Shalev (1996) and Patridge et al (1998) have studied the dosimetric characteristics of photon beams at low MU setting and observed inaccuracy in dose delivery at less than 30 MU. Dosimetric problems at low MU for few scanned and scattering foil electron beams were reported by Das et al (1994). In this study we have analyzed the dosimetric accuracy at low MU setting for electron beams from 6 MeV to 20 MeV at different pulse repetition frequencies.
4.2 MATERIALS & METHODS

4.2.1 Phantom measurements:

The dosimetric accuracy at low MU setting was studied using he dose measurements by a 0.6 cc Farmer type ion chamber (FROB-G) with a Capintec dosimeter (M/S Capintec, USA) at $D_{\text{max}}$ in polystyrene phantom of $25\times25\times30 \text{ cm}^3$. The field size was $10\times10 \text{ m}^2$ at the nominal source to phantom surface distance of 100 cm. The nominal electron beam energies of 6, 9, 12, 16 and 20 MeV available from a high energy Linac (Clinac-1800, M/S Varian, USA) have been used. MU were selected from lowest possible value of 1 MU upto 200 MU at five different pulse repetition frequencies (PRF). The change in PRF resulted in dose rate variation from 80 to 400 MU/minute. The response of ion chamber used in this study was checked with Theratron 780C cobalt teletherapy machine for its accuracy at very low integrated exposures by employing large source to chamber distances beyond 200 cm.

The dose to the medium was estimated as described in Technical report series 277 of IAEA (TRS 277, 1987) and is given by

$$D_{\text{med}} = Q \times N_D \times S_{\text{med, air}} \times P_u$$

where $Q$ is the average measured charge corrected for temperature and pressure, $N_D$ is the absorbed dose factor for air chamber ($N_{\text{gas}}$ of AAPM protocol), $S_{\text{med, air}}$ is the averaged stopping powers for medium and air, $P_u$ is the perturbation factor relating to cavity dimensions.
5.4.2.2 Dose linearity ratios:

Dose linearity ratio (DLR) for the given MU setting \( (M_{\text{Test}}) \) to be tested is defined as

\[
\text{DLR}_{\text{Test}} = \frac{(D_{\text{med}}/M)}{(D_{\text{med}}/M)_{\text{Cal}}}
\]

where the subscript "Test" refers to the MU setting for checking the linearity at low MU level and "Cal" refers to the calibration MU at nominal setting of 200 MU. In an ideal situation DLR should be unity over the range of MU used clinically. Eq. 2 can be rewritten as

\[
\text{DLR}_{\text{Test}} = \frac{(Q/M)_{\text{Test}} * N_D * S_{\text{med,air}} * P_u}{(Q/M)_{\text{Cal}} * N_D * S_{\text{med,air}} * P_u}
\]

Since \( N_D \) and \( P_u \) are dosimeter and air cavity dependent factors they are constant in the above circumstances. Since the energy variation for scattering foil electron beam is minimum (±2%) at low MU setting [6], \( S_{\text{med,air}} \) is considered as constant. Therefore Eq. 3 is written as

\[
\text{DLR}_{\text{Test}} = \frac{(Q/M)_{\text{Test}}}{(Q/M)_{\text{Cal}}}
\]

The measured DLRs are plotted in semi-log graph for various MU settings selected.
Fig. 5.13 Accuracy of dose delivery for 6 MeV electrons at low monitor unit settings for different dose rates.

Fig. 5.14 Accuracy of dose delivery for 9 MeV electrons at low monitor unit settings for different dose rates.
Fig. 5.15 Accuracy of dose delivery for 12 MeV electrons at low monitor unit settings at different dose rates.

Fig. 5.16 Accuracy of dose delivery for 18 MeV electrons at low monitor unit setting for different dose rates.
Fig. 5.17 Accuracy of dose delivery for 20 MeV electrons at low monitor unit settings for different dose rates.

Fig. 5.18 Accuracy of dose delivery at 80 MU/min settings for various electron energies.
Fig. 5.19 Accuracy of dose delivery at 160 MU/min setting for various electron energies.

Fig. 5.20 Accuracy of dose delivery at 320 MU/min setting for various electron energies.
Fig. 5.21 Accuracy of dose delivery at 320 MU/min setting for various electron energies.

Fig. 5.22 Accuracy of dose delivery at 400 MU/min setting for various electron energies.
5.4.3 RESULTS

The dose linearity ratio (DLR) for different MU settings is shown in Fig.5.13 for 6 MeV electrons. The variation in DLR for different dose rates (PRF) are also shown in the same figure. It is found that the DLR is almost constant from 200 MU down to about 10 MU at all the dose rates (PRF). Below 10 MU setting the DLR increases significantly reaching the maximum variation at 1 MU setting for all PRF. The inaccuracy in dose linearity below 10 MU is more as the dose rate increases. Similar results are obtained for 9, 12 and 16 MeV electron beams (Fig.5.15, Fig.5.16 and Fig.5.17). The DLR value is found to be constant down to 10 MU setting, for high electron energy of 20 MeV (Fig.5.14).

For a selected dose rate (PRF) the dose linearity inaccuracy increases with increase in nominal electron beam energy. The variation of DLR for different electron energies at 80 MU/min and 400 MU/min dose rate setting are shown in Fig.5.18 and Fig.5.19. Similar results for 160 MU/min, 240 MU/min and 320 MU/min are shown in Fig.5.20, Fig.5.21 and Fig.5.22. The DLR value does not increase much with beam energy at low dose rate setting (80 MU setting) as shown in Fig.5.18. However the change in DLR is quite significant at higher dose rate setting (400 MU) setting as shown in Fig.5.19.
5.5 MEASUREMENT OF PHYSICAL PARAMETERS FOR TOTAL SKIN ELECTRON THERAPY

5.5.1 SCIENTIFIC BACKGROUND:

Total skin electron therapy (TSET) with energies 2-7 MeV at the patient body surface and 4-10 MeV at the accelerator exit window is one of the most effective treatment for mycosis fungoides (cutaneous T cell lymphoma). The aim of this technique is to deliver sufficiently uniform dose over the entire height of the body surface to the skin with minimum dose to the underlying normal structures. A complex summary of different techniques and requirements for dosimetry of TSET has been published by the American Association of Physicists in Medicine (AAPM 1988). In order to cover the entire body surface, a field size of about 200 cm height and 80 cm width is required, which can be obtained at larger treatment distances than normally encountered from the scattering foil of a medical electron accelerator. With careful planning and execution a dose uniformity of ±8% vertically over the central 160 cm and ±4% horizontally over the central 60 cm can be achieved. Many treatment techniques like narrow rectangular beams (Lo et al 1979, 1981), scattered single beam (Tefenes and Goodwin 1977), two parallel beams (Szur et al 1962), pairs of angled beams (Hoppe et al 1979, Hoppe et al 1979), pendulum arc technique (Sewchand et al 1979) and patient rotation technique (Kumar and Patel 1978, Podgorsak et al 1983, Tefenes and Goodwin 1977) are reported in literature. A combination of six dual...
ields around the patient (one anterior, one posterior, two anterior obliques and two posterior obliques) are commonly used (Karzmark et al 1960, Karzmark 1964, 1968) for the treatment. The various measurements carried out for TSET at our institute are presented.

5.5.2 MATERIALS & METHODS

5.5.2.1 Geometry of TSET irradiation:

The Clinac-1800 6 MeV electron beam, the minimum nominal energy available in the accelerator, with a dose rate set at 400 MU/min aimed laterally was used to treat the patient (Fig.5.23). The treatment room design facilitate to maintain the patient body surface at a distance of 460 cm with sufficient distance left out from the wall to avoid scatter. To obtain the required energy a perspex sheet degrader of 1 cm thickness was specially designed and mounted in a frame, which can be moved easily with the wheels attached to it. To support the patient arms during irradiation nylon straps are attached to the frame work.

5.5.2.2 Assessment of electron beam energy:

The depth ionization measurements for a single horizontal beam were carried out with a PTW parallel plate ionization chamber embedded in a perspex phantom (25x25x30 cm²) and kept at 20 cm behind the degrader. The distance to the midplane of the scatterer from the scattering foil is maintained at 460 cm. From
FIG. 5.23. IRRADIATION GEOMETRY OF SYMMETRICAL DUAL-FIELD TECHNIQUE

1cm PERSPEX SHEET DEGRADER PLACED AT 20CMS IN FRONT OF THE PATIENT.
the depth ionization curve, the most probable energy at phantom surface \( E_{p,0} \) is found out from the range energy relation (ICRU 1984)

\[
E_{p,0} = 1.95 R_p + 0.48 \quad \text{(1)}
\]

where \( R_p \) is the practical range. The mean energy at the treatment plane \( \bar{E}_0 \) is obtained using the relationship (ICRU 1984)

\[
\bar{E}_0 = 2.33 R_{50} \quad \text{(2)}
\]

where \( R_{50} \) is the half value water equivalent depth in cm. The treatment beam transmitted through the perspex absorber gets degraded and spreads out in energy. The mean energy at depth \( Z \) is estimated from the mean entrance energy \( \bar{E}_Z \) using the relation

\[
\bar{E}_Z = \bar{E}_0 \left[ 1 - \left( \frac{z}{R_p} \right) \right] \quad \text{(3)}
\]

5.5.2.3 Electron beam uniformity for large fields:

The horizontal dose profile is measured in air using lateral beam at the treatment distance with a 0.6 cc Farmer type thimble chamber coupled to a RDM-1F (Therados, Sweden) electrometer. The individual beam dose profile is combined side by side with its mirror image, gives a resultant profile (Fig.5.24). The distance between the central axes of the two beams in the vertical plane at the treatment distance gives a measure of the angle between the two fields.
Fig. 5.24 Resultant dose profile from two angulated larger field electron...
5.5.2.4 Calibration of absorbed dose:

The absolute dose at the point of maximum dose was measured using the Co-60 dose calibrated PTW parallel chamber, using IAEA dosimetry protocol (IAEA TRS 277, 1987). The dose at maximum in water \( (D_w) \) is given by

\[
D_w = M_u \times N_D \times S_{w, \text{air}} \times P_u \quad \text{(4)}
\]

where \( M_u \) is the meter reading corrected for temperature and pressure, \( N_D \) is the absorbed dose calibration factor in water, \( S_{w, \text{air}} \) is the ratio of averaged stopping powers of water to air and \( P_u \) is the factor to allow for the presence of cavity and non-water equivalence of chamber.

5.5.2.5 Verification of dose uniformity:

The treatment condition was reproduced using the Alderson Rando phantom for verification of surface dose uniformity. The LiF TL phosphor discs were positioned at different locations over the body as indicated in Fig.5.25. The irradiation geometry is shown in Fig.5.26. The irradiated phosphors were read using a TLD reader (Thelmador-Model 6000, DRP, BARC, Mumbai, India).

5.5.3 RESULTS

The most probable energy at the phantom surface \( (E_{p, o}) \) is estimated to be 3.09 MeV and the mean energy obtained at phantom
Fig. 5.25 Location of TL phosphor on humanoid phantom and the Thelmedor TL reader.

Fig. 5.26 Lateral electron beam directed towards the humanoid phantom positioned behind the perspex degrader.
Fig. 5.27 Variation of relative ionization with depth indicating the practical range and Bremsstrahlung dose from the degraded electron beam.
Fig. 5.28 The surface dose variation for combined fields estimated by LiF TL phosphor.
surface is 2.18 MeV. The mean energy at dose maximum $E_x$ is 2.07 MeV. The bremsstrahlung dose from the single degraded electron beam is within ±2% of the electron dose at maximum (Fig. 5.27).

The combined vertical dose profile showed that the electron beam angled at 8° above and below horizontal gives a dose uniformity of ±6% over 210 cm (Fig. 5.24) which can sufficiently cover the patient's entire body surface. The dose at calibration reference point (the umbilicus level) at the depth of maximum ionization is 37 MU/cGy at 400 MU/min dose rate. The surface dose variations over the anterior surface from head down to thigh of the humanoid phantom for combined fields estimated by LiF TL phosphor shows a range of 93% to 104% with respect to 100% at the umbilical level (Fig. 5.28).