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ISSUES INVOLVED IN ABSORBED DOSE IN RADIOTHERAPY

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Abstract: When a beam of radiation falls on a patient (or simulator), the dose absorbed varies with depth, characterizing a tissue dose distribution human. This variation depends on several parameters: beam energy, depth, medium composition, field size, source distance and collimation system of the beam.
Keywords: Dose absorbed, radiotherapy

DOSE ABSORBED

The absorbed dose D is a fundamental dosimetric quantity defined by equation 1.

$$D = \frac{d\overline{E}}{dm}$$

where Ed is the average energy transferred by ionizing radiation to matter in a volume elementary volume and dm is the mass in that elementary volume. In the International System, the unit of absorbed dose is the joule per kilogram (J/kg), whose name is the gray (Gy).

DOSE DISTRIBUTION

When a beam of radiation falls on a patient (or simulator), the dose absorbed varies with depth, characterizing a tissue dose distribution human. This variation depends on several parameters: beam energy, depth, medium composition, field size, source distance and collimation system of the beam.

In the case of radiotherapy treatments with electron beams, most advantage lies in the shape of the depth dose distribution curve on the axis central, especially in the range of 6 to 15 MeV. This dose distribution curve is characterized by an almost uniform dose region followed by a rapid drop in same. This feature is a clinical advantage over other conventional megavoltage photon treatment modalities.

Because of differences in electron beam generation and collimation, depth dose distribution and surface dose may be different for different clinical treatment accelerators. In clinical practice, however, it is not.

It is sufficient to specify only the beam energy, but several parameters that characterize the dose variation in depth along the central axis of the electron beam. This step is fundamental in the procedure for calculating the dose administered to the patient. This section defines some parameters that allow characterizing the dose distribution in the human body.

PERCENTAGE OF DOSE IN DEPTH

The percent depth dose (PDP) curve is the ratio between the absorbed dose at a given depth and the absorbed dose at a reference depth, usually the maximum dose dept, for example:.the depth of electronic equilibrium, as described in equation 2. The region where the absorbed dose increases until it reaches the maximum value is called the build-up region. The PDP depends on the energy, depth, field size, scattering, and source-to-surface distance.

$$PDP_d(d, f, C, h\nu) = \frac{D_x}{D_{Max}}.100$$

where xD is the dose at the depth in question, MaxD is the dose at the maximum dose depth and dPDP is the percentage of depth dose at depth d. The PDP depends on depth d, source type f, field size C and beam energy HV. Typical PDP in water curves for clinical electron beams for field size 10x10 cm 2 on the central axis are illustrated in Figure 1.

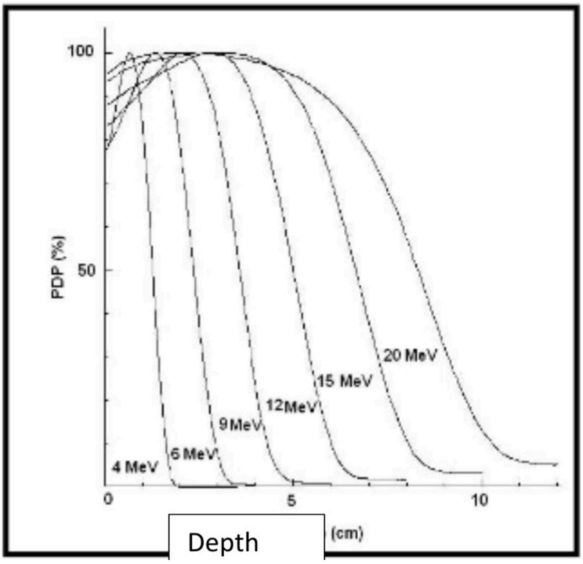

Figure 1: Typical PDP curves for clinical electron beams for field size 10x10 cm 2 on the central axis in water [1].

Typically, the depth dose curve on the central axis of an electron beam exhibits a high surface dose compared to the depth dose curve for a photon beam illustrated infFigure 2. In the case of electrons, the depth dose curve remains more or less uniform and drops rapidly with depth. These features offer a clinical advantage over clinical photon beams in the treatment of superficial tumors.

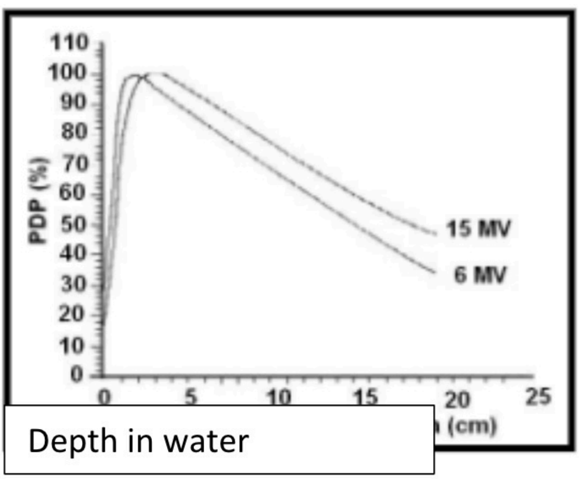

Figure 2: Depth dose curves for a photon beam from a clinical linear accelerator for field size 10x10 cm 2 for water [1].

The depth dose for a specific energy will be essentially regardless of field size when the distance between the central axis and the field edge is greater than the lateral range of the scattered electrons. In this case, lateral electronic equilibrium exists. As the field size decreases, there is no electronic equilibrium on the central axis and the depth dose will be very sensitive to the field size, as can be seen in Figure 3 for an electron beam of 20 MeV and field sizes smaller than 25x25 cm 2.

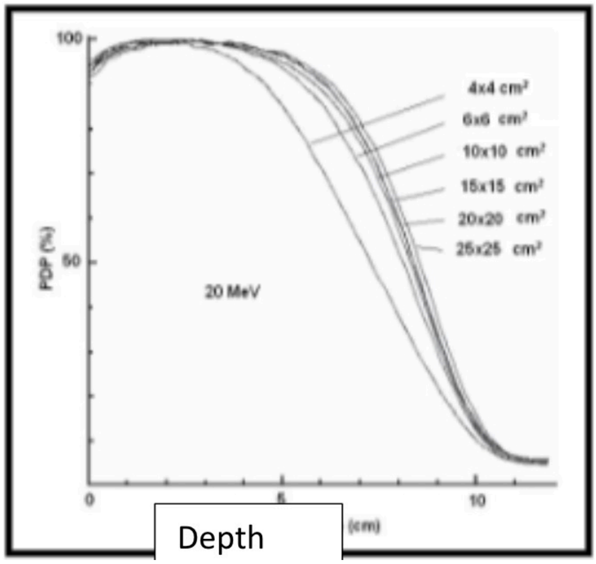

Figure 3: Comparison of depth dose profiles in water of central axis electron beams for 20 MeV energy and various field sizes [1].

The distributions illustrated in figures 2 and 3 are given for the perpendicular incidence of the electron beam on the patient or simulator. For oblique views with arbitrary angles α between the central axis and the patient or simulator, there are significant changes in the characteristics of the PDP curves for the electron beam. For small angles of incidence α, the slope of the PDP decreases and the practical range is essentially invariant with respect to the normal incidence. When the angle of incidence exceeds 60°, the PDP loses its characteristic shape and range definition practical. Therefore, this PDP

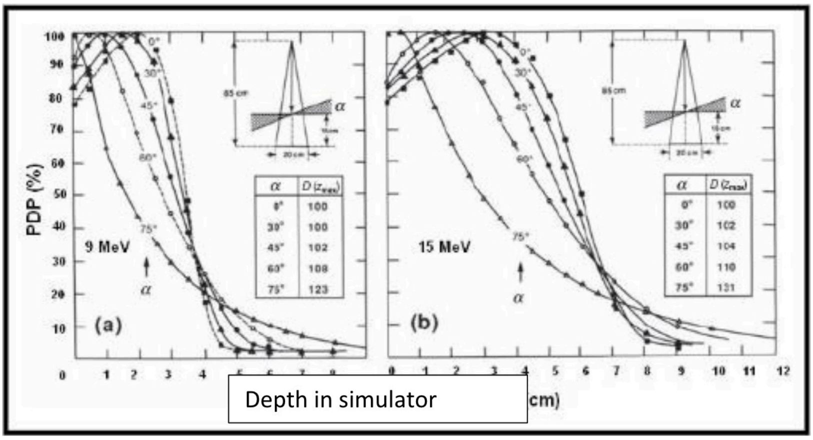

Figure 4: PDP curves in water for various electron beam incidences. a) 9 MeV and b) 15 MeV. The normal incidence is represented by α=0°. In the detail of the figure, the maximum dose depth (MaxD) for various angles is highlighted [1].

cannot be applied. Figure 4 shows the effect of the angle of incidence α on the absorbed dose distribution [1]. Oblique views are used in certain clinical situations where it is necessary for the electron beam to cross curved regions.

From the depth dose percentage curve, it is possible to determine other parameters necessary for the characterization of the energy of the clinical electron beam. The five definitions of the different range parameters for a central axis depth dose distribution are illustrated in figure 5.

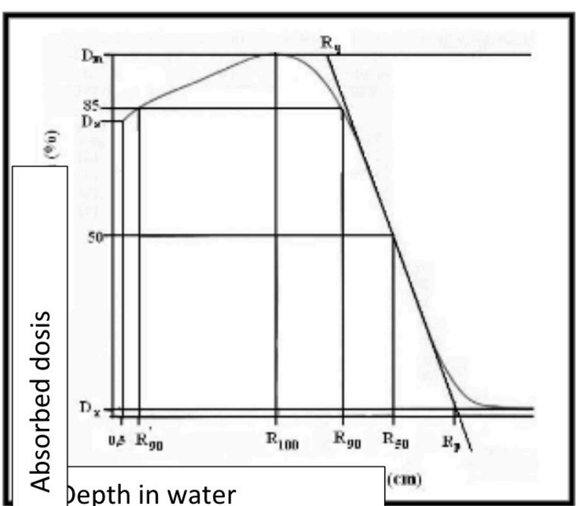

Figure 5: Typical PDP curve illustrating the parameters: R_q, R_{90}, R_{50}, R_p, R_{max}.

Three of these parameters are range parameters (R 90, R 50, R p), and two are measurements at the input (D s) and output (D x) dose levels in the irradiated volume.

CLINICAL PARAMETERS OF THE ELECTRON BEAM

Due to the complexity of the spectrum, there is no single energy that completely characterizes an electron beam. Various parameters are used to describe a beam such as the probable energy (E p,0), the average energy on the surface of the geometric simulator or simulator, ($\overline{E}_0$)and R50.

Practical scope

The practical range (Rp) is defined as the depth where the tangent at the point of greatest slope of the PDP curve (inflection point) intercepts the radiation background (D x). This value, Rp, is obtained by making a linear fit of the data in the descending region of the depth dose curve between 70% and 20%. The photon background, which is defined as the extrapolation from the end of the dose distribution in depth to the practical range, is also obtained by making a linear fit on the values obtained up to the last point of the dose distribution curve [2].

Due to the complexity of the electron spectrum generated by the linear accelerator, the energy calculated from the practical range is more directly related to the most likely energy of the electron beam when it reaches the surface of the simulator geometric. Equation 3 empirically relates the most likely energy of the electron beam to the practical range [1].

$$E_{p,0} = 0,22 + 1,09R_p + 0,0025\, R^2_p$$

Mid-Depth Reach

The mid-depth range (R 50) is the depth at which the absorbed dose reaches 50% of its maximum value. In electron dosimetry, the clinical beam energy is characterized by the R 50, also called the electron beam quality index, as specified in TRS 398 [4]. This index denotes the average energy of the electron beam, since the beam originating from the accelerator has a complex spectrum. As the half depth is located in the region reached by approximately half of the primary electrons, it is reasonable to assume that this depth is more related to the average energy than to the most likely energy of the incident electrons. The average energy of electrons, 0E, on the surface of the geometric simulator or the patient is related to R 50 as shown in relation (4):

$$\overline{E}_0 = CR_{50}$$

where C = 2.33 MeV/cm for water.

In figure 6, the variation of R 50 with the average energy on the surface of the simulator, $\overline{E}_0$.[2]

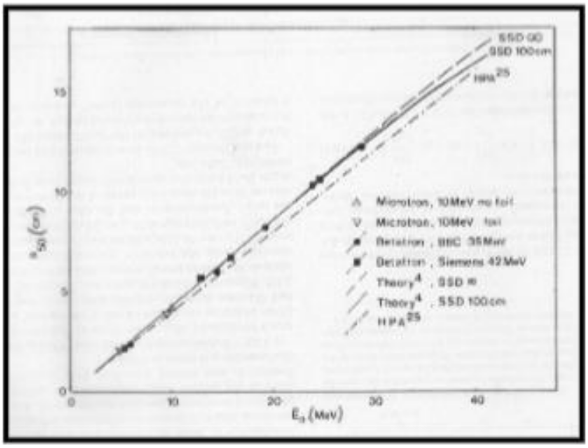

Figure 6: Variation of R 50 with the average energy of several research accelerators [2].

Therapeutic Reach

The therapeutic range (R 90) is the depth at which the absorbed dose reaches 90% of its maximum value. This, if possible, must coincide with the distal margin of the treatment. For a given energy, the length of this interval depends on the desired uniformity of dose distribution in depth in the target region. A dose difference of less than or greater than 10% can cause a considerable change in the ratio between the fraction of patients cured and those who will experience complications, a fact that makes the effort to obtain a small variation within the target volume desirable [3].

Superficial dose

The entry dose in a volume irradiated by a clinical electron beam is of great interest because of its importance for the degree of spared skin obtained. The depth of the radiation-sensitive layers below the epidermis is generally order of 0.5 mm, a depth that is accessible for accurate dose measurements with multiple detectors. The surface dose (D s) is defined as the ratio of the absorbed dose at a depth of 0.5 mm in relation to the maximum dose value.

Photon background radiation or X-ray contamination

Photon background radiation or X-ray contamination in the electron beam (D x) is defined as the extrapolation from the tail of the dose distribution in depth to the practical range. Photons are present as a contamination of the incident electron beam or are generated in the irradiated medium itself.

FIELD PROFILE

Dose distributions along the central axis provide only part of the information required for accurate dose description within the patient.

Two-dimensional dose distributions are determined with central axis information in conjunction with field dose profiles.

Field profiles are obtained by measuring the dose perpendicular to the central axis at an arbitrary depth along the measurement geometric simulator.

The usual measurement depths are MaxD and 10 cm in order to check compliance with the machine's specifications. Other depths are normally required to configure the planning system. An example of a field profile measured at the maximum dose depth is shown in Figure 7.

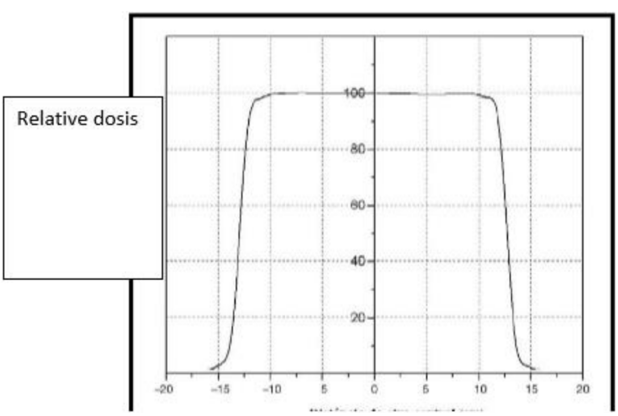

Figure 7: Example of a field profile for an electron beam obtained from a clinical accelerator with energy of 12 MeV, maximum dose depth and field size of 25x25 cm 2.

The field profile is obtained through normalization with the dose on the central axis according to equation 5.

$$D_R(D_x, f, C, h\nu) = \frac{D_i}{D_z}.100$$

where RD is the dose relative to the central axis, zD is the dose at a certain depth on the central axis, ie at 0 on the X axis. iD is the dose on the X axis. The Y axis can also be used to obtain a profile field.

ISODOSE CURVES

The dose on the central axis is not always sufficient for planning the treatment of the patient and often information about the complete isodose curve is required.

Isodose curves are lines passing through points of the same dose. These curves are obtained at regular absorbed dose intervals and are expressed as the percentage dose relative to a reference point, normally the maximum dose point on the central axis. As a beam of electrons penetrates a medium, it becomes expands rapidly below the surface due to scattering. However, the individual scattering of the isodose curves varies with the level of isodose, energy of the beam, field size and beam collimation [3].

Isodose curves for a 20 MeV electron beam for the 10x10 cm 2 field size and 100 cm SSD are illustrated in Figure 8.

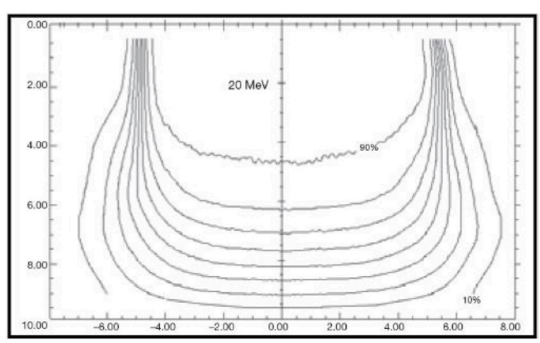

Figure 8: Isodose curves for energy 20 MeV and field size 10x10 cm 2 and 100 cm SSD.

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It is not appropriate to measure dose distribution directly in radiation-treated patients. Dose distribution information is entirely measured in simulators, usually with sufficient volume to provide attenuation and scattering conditions similar to the human body.

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