Dose distribution and scatter analysis

It is seldom possible to measure dose distribution directly in patients treated with radiation. Data on dose distribution are almost entirely derived from measurements in phantoms tissue equivalent materials, usually large enough in volume to provide full-scatter conditions for the given beam. These basic data are used in a dose calculation system devised to predict dose distribution in an actual patient.

PHANTOMS

Basic dose distribution data are usually measured in a water phantom, which closely approximates the radiation absorption and scattering properties of muscle and other soft tissues.

Another reason for the choice of water as a phantom material is that it is universally available with reproducible radiation properties.

A water phantom, however, poses some practical problems when used in conjunction with ion chambers and other detectors that are affected by water, unless they are designed to be waterproof. In most cases, however, the detector is encased in a thin plastic (water equivalent) sleeve before immersion into the water phantom.

Since it is not always possible to put radiation detectors in water, solid dry phantoms have been developed as substitutes for water. Ideally, for a given material to be tissue or water equivalent, it must have the:

1. Same effective atomic number
2. Same number of electrons per gram
3. Same mass density.

Anthropomorphic phantoms are frequently used for clinical dosimetry. One such commercially available system, known as Alderson Rando Phantom,' incorporates materials to simulate various body tissues muscle, bone, lung, and air cavities. The phantom is shaped into a human torso and is sectioned transversely into slices for dosimetric applications.

DEPTH DOSE DISTRIBUTION

- ❖ As the beam is incident on a patient (or a phantom), the absorbed dose in the patient varies with **depth**.
- ❖ This variation depends on many conditions:
  beam energy, depth, field size, distance from source, and beam collimation system.

Thus the calculation of dose in the patient involves considerations in regard to these parameters and others as they affect depth dose distribution.

An essential step in the dose calculation system is to establish depth dose variation along the central axis of the beam.

A number of quantities have been defined for this purpose, major among these being percentage depth dose tissue-air ratios, tissue-phantom ratios, and tissue-maximum ratios. These quantities are usually derived from measurements made in water phantoms using small ionization chambers.

Although other dosimetry systems such as TLD, diodes, and film are occasionally used, ion chambers are preferred because of their better precision and smaller energy dependence.

PERCENTAGE DEPTH DOSE

One way of characterizing the central axis dose distribution is to normalize dose at depth with respect to dose at a reference depth. The quantity percentage (or simply percent) depth dose may be defined as the quotient, expressed as a percentage, of the absorbed dose at any depth **d** to the absorbed dose at a fixed reference depth do, along the central axis of the beam. Percentage depth dose (P) is thus:

$$P = \frac{D_d}{D_{d_0}} \times 100$$

For orthovoltage (up to about 400 kVp) and lower-energy x-rays, the reference depth is usually the surface ($d_o = 0$). For higher energies, the reference depth is taken at the position of the peak absorbed dose ($d_o = d_m$).

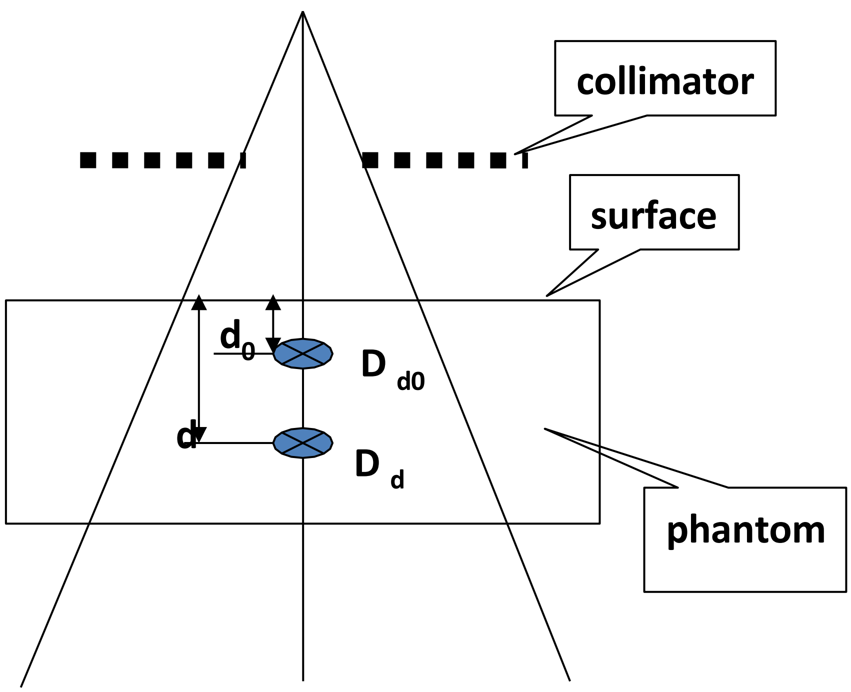


FIG.1 Percentage depth dose is *(D_dlDd_o)* x 100, where *d* is any depth and d_o is: reference depth of maximum dose.

Email: ahed.hameed.jaaz@uomus.edu.iq

In clinical practice, the peak absorbed dose on the central axis is sometimes called the *maximum dose*, the *dose maximum*, the *given dose*, or simply the D_m,. Thus,

$$D_{\max} = \frac{D_d}{P} \times 100$$

A number of parameters affect the central axis depth dose distribution. These include :

1. Beam quality or energy
2. Field size and shape,
3. Source to surface distance
4. Beam collimation.

Dependence on Beam Quality and Depth

The percentage depth dose (beyond the depth of maximum dose) increases with beam energy. Higher-energy beams have greater penetrating power and thus deliver a higher percentage depth dose.

The physics of dose buildup may be explained as follows:

(a) As the high-energy photon beam enters the patient or the phantom, high-speed electrons are ejected from the surface and the subsequent layers

(b) These electrons deposit their energy a significant distance away from their site of origin

(c) Because of (a) and (b), the electron fluence and hence the absorbed dose increases with depth until they reach a maximum.

Email: ahed.hameed.jaaz@uomus.edu.iq

- ❖ However, the photon energy fluence continuously decreases with depth and, as a result, the production of electrons also decreases with depth.
- ❖ The net effect is that beyond a certain depth the dose eventually begins to decrease with depth.
- ❖ It may be instructive to explain the buildup phenomenon in terms of absorbed dose and a quantity known as kerma (from kinetic energy released in the medium).

*** the kerma (K) defined as "the quotient of dE,, by dm, where dE,, is the sum of the initial kinetic energies of all the charged ionizing particles (electrons) liberated by uncharged ionizing particles (photons) in a material of mass dm.

$$K = \frac{dE_{tr}}{dm}$$

- ❖ Because kerma represents the energy transferred from photons to directly ionizing electrons, the kerma is maximum at the surface and decreases with depth because of the decrease in the photon energy fluence.
- ❖ The absorbed dose, on the other hand, first increases with depth as the high-speed electrons ejected at various depths travel downstream. As a result, there is an electronic build-up with depth.
- ❖ However, as the dose depends on the electron fluence, it reaches a maximum at a depth approximately equal to the range of electrons in the medium. Beyond this depth, the dose decreases as kerma continues to decrease, resulting in a decrease in secondary electron production and hence a net decrease in electron fluence.
- ❖ As shown in figure below the kerma curve is initially higher than the dose curve but falls below the dose curve beyond the build-up region. This effect is explained by the fact that the areas under the two curves taken to infinity must be the same.

Email: ahed.hameed.jaaz@uomus.edu.iq

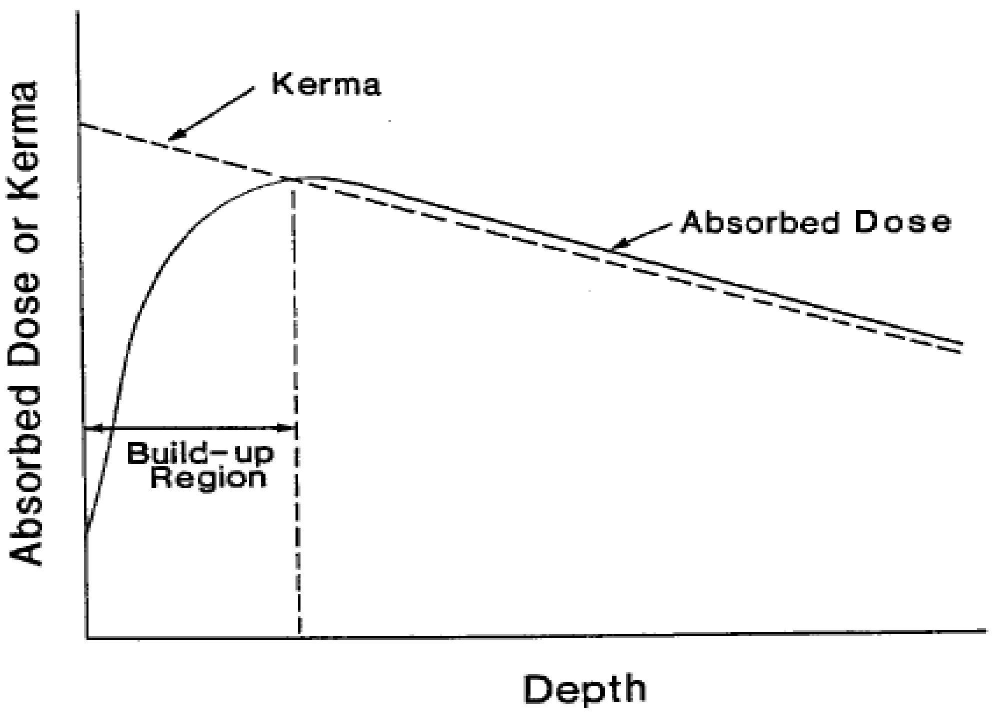


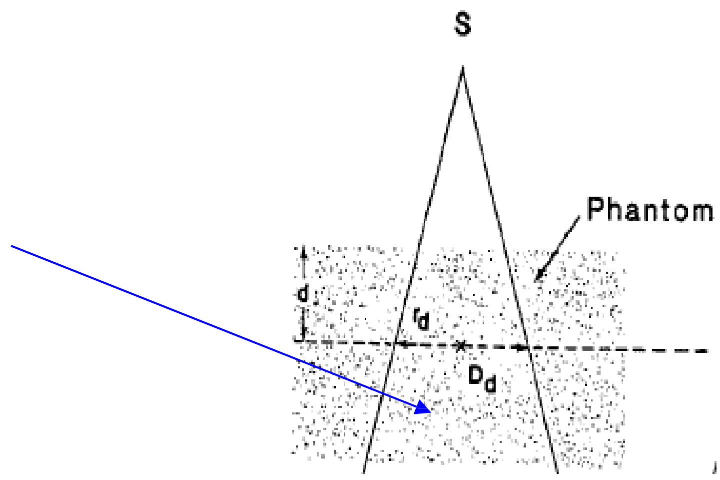
Figure: Schematic plot of absorbed dose and kerrna as function of depth.

Effect of Field Size and Shape

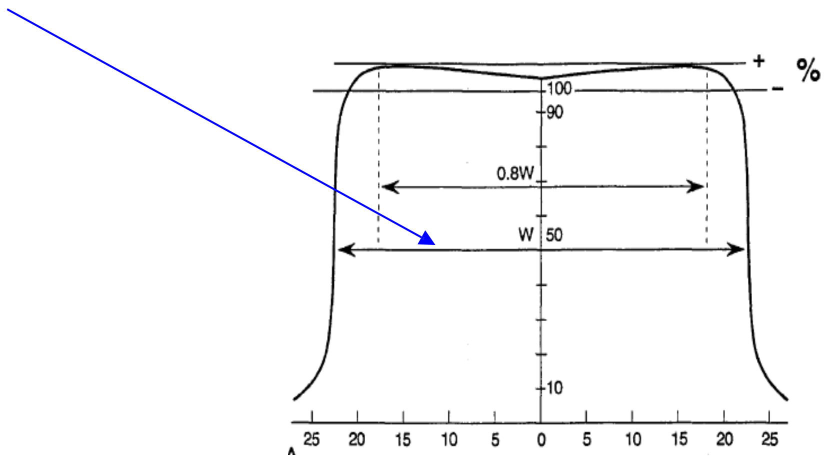
- Field size may be specified either geometrically or dosimetrically. **The geometrical field size** is defined as "the projection, on a plane perpendicular to the beam axis, of the distal end of the collimator as seen from the front center of the source".
- This definition usually corresponds to the field defined by the light localizer, arranged as if a point source of light were located at the center of the front surface of the radiation source.
- The **dosimetric or the physical**, field size is the distance intercepted by a given isodose curve (usually 50% isodose) on a plane perpendicular to the beam axis at a stated distance from the source.

- In addition, the field size will be defined at a predetermined distance such as the source surface distance (SSD) or the source-axis distance (SAD). The latter term is the distance from the source to axis of gantry rotation known as the isocenter.
- For a sufficiently small field one may assume that the depth dose at a point is effectively the result of the primary radiation, that is, the photons which have traversed the overlying medium without interacting. The contribution of the scattered photons to the depth dose in this case is negligibly small or 0.
- But as the field size is increased, the contribution of the scattered radiation to the absorbed dose increases. Because this increase in scattered dose is greater at larger depths than at the depth of D_m, the percent depth dose increases with increasing field size
- The increase in percent depth dose caused by increase in field size depends on beam quality. Since the scattering probability or cross-section decreases with energy increase and the higher-energy photons are scattered more predominantly in the forward direction, the field size dependence of percent depth dose is less pronounced for the higher-energy than for the lower-energy beams.
- Percent depth dose data for radiation therapy beams are usually tabulated for square fields. Since the majority of the treatments encountered in clinical practice require rectangular and irregularly shaped (blocked) fields, a system of equating square fields to different field shapes is required.

Geometrical



Dosimetrically or *physical*



Dependence on Source-Surface Distance:

- Photon fluence emitted by a point source of radiation varies inversely as a square of the distance from the source
- The actual dose rate at a point decreases with increase in distance from the source, the percent depth dose, which is a relative dose, increases with SSD
- Although the actual dose rate at a point decreases with increase in distance from the source, the percent depth dose, which is a relative dose with respect to a reference point, increases with SSD
- In clinical radiation therapy, SSD is avery important parameter. Because percent depth dose determines how much dose can be delivered at depth relative to the surface dose or Dm, the SSD needs to be as large as possible.
- However, because dose rate decreases with distance, the SSD, in practice, is set at a distance which provides a compromise between dose rate and percent depth dose. For the treatment of deep-seated lesions with megavoltage beams, the minimum recommended SSD is 80 cm.

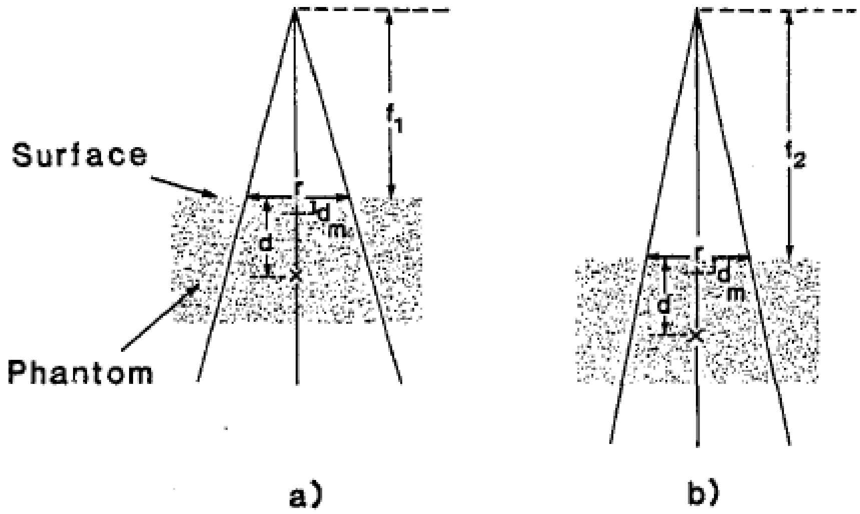


Fig: Change of percent depth dose with **SSD.** Irradiation condition (a) has **SSD** = ***f1*** and condition **(b)** has **SSD** = ***f2.*** For both conditions, field size on the phantom surface, ***r*** **x** *r_1* and depth d are the same.

Email: ahed.hameed.jaaz@uomus.edu.iq