Relative dosimetry: output factors, profiles, penumbra and depth functions

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Introduction

The dose deposition in a patient is a very complicated process.
It’s must take in account the **attenuation** and **scattering** of the photon beam inside a large and various volume.

Data on dose distribution in patients is derived from measurements in **tissue-equivalent-phantoms** large enough to provide full scatter conditions.

Several empirical functions are used to link the dose at any arbitrary point inside the patient/phantom to the known dose at the reference point in a phantom.
Introduction
Dosimetric functions are measured in tissue equivalent phantoms with suitable radiation detectors. Dosimetric functions are determined for a specific set of reference conditions:

- **Depth z**
- **Field Size**
- **Source-Surface Distance (SSD) or Source-Axis Distance (SAD)**

There are two types of data:

1) **scanned data**
2) **non-scanned data or point dose data**

**Scanned beam data collection** is carried out with a scanning **water phantom**; typically, a plastic tank filled with water to a level deep enough to allow central axis PDD and profile measurements to a depth of 40 cm.

**Point dose data** can be measured in a **solid phantom** or in a **water phantom**.
Dosimetric data

✓ **Central axis depth dose at standard SSD set-up:**
  ✓ **PDD**

✓ **Central axis depth dose at standard SAD set-up:**
  ✓ Tissue Air Ratio (TAR)
  ✓ Tissue Phantom Ratio (TPR)
  ✓ Tissue Maximum Ratio (TMR)

✓ **Total scatter factor** $S_{cp}$
✓ **In-air output ratio** $S_{c}$
✓ **Phantom scatter factor** $S_{p}$
✓ **Beam profiles, penumbra and off axis factors**
Phantoms

Water phantom closely approximates the radiation absorption and scattering properties of muscle and soft tissues. Main dosimetrical data are measured in water but for particular conditions it’s not possible and solid water-equivalent phantom were developed.

The electron density $\rho_e$ of material must be equal to water $\rho_e$:

$$\rho_e = \rho_m N_A \frac{Z}{A}$$
water phantom

To perform isodose measurement in water with different type of ionization chamber, diodes.

Software dedicated to evaluate parameters of beams
water phantom

The size of the water tank should be large enough:

- to allow scanning of beam profiles up to the largest field size required (e.g., for photon beams, 40x40 cm$^2$ with sufficient lateral buildup 5 cm and overscan distance)

- to allow larger lateral scans and diagonal profiles for the largest field size and at a depth of 40 cm for modeling as required by some planning systems


_to determine the appropriate size of the scanning tank, the overscan and the beam divergence at 40 cm depth should be considered._
Solid water-water equivalent Phantom

Water equivalent phantom with (a) Farmer-type ion chamber and (b) parallel-plate chamber

The solid plate phantom (PMMA) may be used for dosimetry measurements in photon and electron beams, based on the relation between ionization chamber reading in plastic and water in the user beam with different types of ionization chambers.
Percent depth dose PDD

For indirectly ionizing radiations, energy is imparted to matter in a two step process:
1) the indirectly ionizing radiation transfers energy as kinetic energy to secondary charged particles (kerma).
2) These charged particles transfer some of their kinetic energy to the medium (absorbed dose) and lose some of their energy in the form of radiative losses.

**Kerma** (kinetic energy released per unit mass) is defined as the mean energy transferred from the indirectly ionizing radiation to charged particles (electrons) in the medium per unit mass dm:

\[ K = \frac{d\bar{E}_{tr}}{dm} \]

The **absorbed dose** \( D \) is defined as the mean energy \( \varepsilon \) imparted by ionizing radiation to matter of mass \( m \) in a finite volume \( V \) by:

\[ D = \frac{d\bar{\varepsilon}}{dm} \]
Percent depth dose PDD

The dose at point Q in the patient consists in two component:

*primary component* and *scatter component*

\[
PDD(z, A, f, h\nu) = \left(\frac{f+z_{max}}{f+z}\right)^2 \cdot e^{-\mu_{eff}(z-z_{max})} \cdot Ks
\]

Ks is the scattering component.

This indicates the three governing rules of photon beam attenuation: *inverse square law, exponential attenuation* and *scattering component*.

Percent Depth Dose uniquely varies with depth due to attenuation, with SSD due to inverse square law, and with field size due to scattering effect.

- The *primary component* is the photon contribution to the dose at point Q that arrives directly from the source.

- The *scatter dose* is delivered by photons produced through Compton scattering in the patient, machine collimator, flattening filter or air.
**Percent depth dose PDD**

The *percentage depth dose* is defined as the quotient of the absorbed dose at any depth \( d \) to the absorbed dose at a fixed reference depth \( d_0 \) along the central axis of the beam:

\[
P = \frac{D_d}{D_{d0}} \times 100
\]

For high energies the reference dose is taken at the position of the *peak absorbed dose*.
Percent depth dose PDD

As the beam is incident on a phantom (as on a patient) the absorbed dose varies with depth.

This variation depends on many conditions:

- **beam energy** \((h\nu)\)
- **Depth** \((z)\)
- **field size** \((A)\)
- **distance from source** \((SSD)\)
- **beam collimation system.**

\[
PDD (z, A, f, h\nu) = \frac{D_Q}{D_P} \times 100
\]
**Percent depth dose PDD: dependence on depth**

The percentage depth dose (PDD) for a constant $A$, $f$ and $hv$ first increases from the surface to $z = z_{\text{max}}$ (*build-up region*) and then decreases with $z$. 

![Graph showing the build-up, skin dose, and exponential decay regions of the PDD for a constant A, f, and hv.](image)
Surface dose and build-up region

The dose region between the surface and depth \( z = z_{\text{max}} \) in megavoltage photon beams is referred to as the dose build-up region and results from the relatively long range of energetic secondary charged particles that first are released in the patient by photon interactions (photoelectric effect, Compton effect, pair production) and then deposit their kinetic energy in the patient.

- The depth of dose maximum \( z_{\text{max}} \) beneath the patient’s surface depends on the beam energy and beam field size.
- The beam energy dependence is the main effect
- The field size dependence is often ignored because it represents only a minor effect.
The surface dose represents contributions to the dose from:

(1) Photons scattered from the collimators, flattening filter and air;

(2) Photons backscattered from the patient;

(3) High-energy electrons produced by photon interactions in air and any shielding structures in the vicinity of the patient.

The low surface dose compared to the maximum dose is referred to as the skin sparing effect.
The percentage depth dose increases with beam energy. Higher energy have greater penetrating power.
Percent depth dose PDD: dependence on energy

The percentage depth dose increases with beam energy. Higher energy have greater penetrating power.
Percent depth dose PDD: dependence on energy

The percentage depth dose increases with beam energy. Higher energy have greater penetrating power.
Percent depth dose PDD: dependence on field size

**Geometrical field size**

It’s 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.

**Dosimetric field size**

It’s defined as the distance intercepted by a given isodose curve (usually 50% isodose) on a plane perpendicular to the beam axis at a stated distance from a source (100cm).

- As the field size increases the contribution of scattered radiation to the absorbed dose increases.

- The field size dependence of PDD is less pronounced for the higher energy beams than for the lower energy beams.
Percent depth dose PDD: dependence on field size

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<th>Field size (cm x cm)</th>
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<th>Tolerance (mm)</th>
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![Graph showing percent depth dose (PDD) dependence on field size](image)
In clinical practice a system of equating square field to different filed shapes (typically square field) is required.
Percent depth dose PDD: dependence on SSD

The percentage depth dose (PDD) increases with SSD due to the effects of inverse square law.

The plot shows that the drop in doserate between two points is much greater at smaller distances from the source than at large distance.
Tissue Air Ratio (TAR) is the ratio of the absorbed dose at a given depth in tissue (phantom/patient) to the absorbed dose at the same point in air:

\[
\text{TAR}(z, A_Q, h) = \frac{D_Q}{D'_Q}
\]

- TAR increases with the Beam energy
- TAR increases with the Field size
- TAR decreases with the Depth
- TAR is independent from SSD
Tissue Air Ratio TAR and PDD

\[ \text{TAR}(z, A_Q, h) = \frac{D_Q}{D'_Q} \]

\[ \text{PDD}(z, A, f, h) = 100 \frac{D_Q}{D_P} \]

\[ D_Q = D_P \frac{\text{PDD}(z, A, f, h)}{100} = D'_Q \text{TAR}(z, A_Q, h) \]

\[ \frac{\text{PDD}(z, A, f_1, h)}{\text{PDD}(z, A, f_2, h)} = \left( \frac{\text{TAR}(z, A_{Q1}, h)}{\text{TAR}(z, A_{Q2}, h)} \right)^2 \left( \frac{f_1 + z_{\text{max}}}{f_1 + z} \right) \left( \frac{f_2 + z_{\text{max}}}{f_2 + z} \right) \]
**Pick Scatter Factor (PSF)**

In a phantom the ratio of the dose maximum to the dose in air at the same depth is called pickscatter factor (PSF)

1. PSF increases as the field size increases
2. PSF decreases as the energy increases
3. PSF is independent of SSD
4. PSF increases with field size from unity linearly then saturates at very large field
The tissue phantom ratio TPR is defined as the ratio of the dose at a given point in phantom to the dose at the same point at a fixed reference depth:

$$TPR(z, A_Q, h) = \frac{D_Q}{D_{Q\text{ref}}}$$

$$TMR(z, A_Q, h) = \frac{D_Q}{D_{Q\text{max}}}$$

TPR and TMR depend on the three parameters: $z, A_Q, h$

- $A_Q$ and $h$ constant TMR decreases with increasing $z$.
- $z$ and $h$ constant TMR increases with increasing $A_Q$.
- $z$ and $A_Q$ constant TMR increases with increasing $h$.
Tissue Maximum Ratio TMR and PDD

\[
\text{TMR}(z, A_Q, \nu) = \frac{D_Q}{D_{Q\text{max}}} \\
\text{PDD}(z, A, f, \nu) = 100 \frac{D_Q}{D_P} \\
\text{TMR}(z, A_Q, \nu) \approx \frac{\text{PDD}(z, A, f, \nu)}{100} \left( \frac{f + z}{f + z_{\text{max}}} \right)^2
\]

\[
D_Q = D_P \frac{\text{PDD}(z, A, f, \nu)}{100} = D_{Q\text{max}} \text{TMR}(z, A_Q, \nu)
\]
Collimator scatter correction factor ($S_c$) or Output factor

Collimator scatter correction Factor ($S_c$) is commonly called the Output factor.

It’s defined as the ratio of the output in air for a given field to that for a reference field (e.g. 10x10cm$^2$)

$S_c$ may be measured with an ion chamber with a build cap of size large enough to provide maximum dose buildup for the given energy beam.

Normally $S_c$ are measured at the SAD
Phantom scatter correction factor $S_p$ and total scatter correction factor $S_{cp}$

The **phantom scatter factor** $S_p$ is as the ratio of dose for a given field size at a reference depth to the dose at the same depth for the reference field size $10 \times 10 \text{ cm}^2$. The phantom scatter describes the influence of the scatter originating in the phantom only.

The **total scatter factor** $S_{cp}$ is defined as the ratio of $D_p(z_{\text{max}}, A, f, h, v)$, the dose at $P$ in a phantom for field $A$, to $D_p(z_{\text{max}}, 10, f, h, v)$, the dose at $P$ in a phantom for a $10 \times 10 \text{ cm}^2$ field.

$$S_{c,p} (A, h, v) = \frac{D_p(z_{\text{max}}, A, f, h, v)}{D_p(z_{\text{max}}, 10, f, h, v)}$$

Measurement Set-up of $S_c$ (a) and $S_{cp}$ (b)
**Phantom scatter correction factor** $S_p$ **and total scatter correction factor** $S_{cp}$

$S_p$ is derived from the total scatter correction factor $S_{cp}$, as the ratio between $S_{cp}$ and $S_c$:

$$S_p(s) \approx \frac{S_{cp}}{S_c}.$$
Wedge transmission factor WF

The **wedge transmission factor** (WF) or **wedge factor** is defined as the ratio of the outputs for a given field size (FS), at the reference depth \( d_{\text{ref}}(d) \), in a full scatter phantom at standard geometry, with and without the presence of a wedge filter:

\[
WF(FS, d) = \frac{D_w(FS, d)}{D_0(FS, d)}
\]
### Wedge transmission factor WF

#### Motorized wedge: Nominal wedge angle 60°

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<tr>
<th>Energy</th>
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<th>direction</th>
<th>Measured wedge factor</th>
<th>Reference wedge factor</th>
<th>D(%) &lt; 2%</th>
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#### Physical wedge

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#### Enhanced Dynamic Wedges

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The chamber positioning is very critical.

The axis of chamber must be perpendicular to the direction of wedge.
Off-axis ratios and beam profiles

Dose distributions in 2-D and 3-D are determined with central axis data in conjunction with off-axis dose profiles. The off-axis data are given with beam profiles measured perpendicularly to the beam central axis at a given depth in a phantom.

The depths of measurement are typically at $z_{\text{max}}$ and 10 cm for verification of compliance with machine specifications, in addition to other depths required by the particular treatment planning system (TPS).

The **off-axis ratio (OAR)** is usually defined as the ratio of dose at an off-axis point to the dose on the central beam axis at the same depth in a phantom.
The field flatness changes with depth.
This is attributed to an increase in scatter to primary dose ratio with increasing depth and decreasing incident photon energy off axis.
beam profiles at different depths 
(10x10 and 30x30)

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<th>DOA</th>
<th>Field Size</th>
<th>Pan. Left</th>
<th>Pan. Right</th>
<th>Beam Size</th>
<th>Profiles</th>
<th>Symmetry</th>
<th>Wedge Angle</th>
<th>Field Size at SID</th>
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beam profiles with wedge
Megavoltage X ray beam profiles consist of three distinct regions:

- **Central**
- **Penumbra**
- **Umbra**
Beam profiles: central region

The **central region** represents the central portion of the profile extending from the beam central axis to within 1–1.5 cm from the geometric field edges of the beam.

The **central region** is affected by the energy of electrons striking the thick target, by the target atomic number and by the flattening filter atomic number and geometric shape.
Beam profiles: penumbral region

In the penumbral region of the dose profile the dose changes rapidly and depends also on the field defining collimators, the finite size of the focal spot (source size) and the lateral electronic disequilibrium.

The dose falloff around the geometric beam edge is sigmoid in shape and extends under the collimator jaws into the penumbral tail region, where there is a small component of dose due to the transmission through the collimator jaws (*transmission penumbra*), a component attributed to finite source size (*geometric penumbra*) and a significant component due to in-patient X ray scatter (*scatter penumbra*).
Beam profiles: penumbral and umbra region

The **physical penumbra** is the sum of the three individual penumbras: *transmission, geometric and scatter*.

The physical penumbra depends on:

- **beam energy**,  
- **source size**,  
- **SSD**,  
- **source to collimator distance**  
- **depth in a phantom**

**Umbra** is the region outside the radiation field, far removed from the field edges and results from radiation transmitted through the collimator and head shielding.
Beam profiles: flatness and symmetry

Dose profile uniformity is measured by a scan along the centre of both major beam axes for various depths in a water phantom.

Two parameters quantify the field uniformity:

- **field (beam) flatness**
- **field (beam) symmetry**
Beam profiles: flatness

The beam flatness $F$ is assessed by finding the maximum $D_{\text{max}}$ and minimum $D_{\text{min}}$ dose point values on the beam profile within the central 80% of the beam width:

$$F = 100 \times \frac{D_{\text{max}} - D_{\text{min}}}{D_{\text{max}} + D_{\text{min}}}$$
Beam profiles: symmetry

A typical symmetry specification is that any two dose points on a beam profile, equidistant from the central axis point, are within 2% of each other.

Alternately, areas under beam profile on each side (left and right) of the central axis extending to the 50% dose level (normalized to 100% at the central axis point) are determined.

**Symmetry** $S$ is calculated from:

$$S = 100 \times \frac{\text{area}_{\text{left}} - \text{area}_{\text{right}}}{\text{area}_{\text{left}} + \text{area}_{\text{right}}}$$
Dose profile measurements
# Dose profile measurements

## Profile parameters (ELEKTA protocol)

<table>
<thead>
<tr>
<th>Energy</th>
<th>FIELD</th>
<th>Axis</th>
<th>FIELD Size(mm)</th>
<th>Flatness(%)</th>
<th>Symmetry(%)</th>
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## Profile parameters (home protocol)

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Isodose curves

In order to represent volumetric and planar variation in absorbed dose, distribution are depicted by means of ISODOSE CURVES

Isodose curve are the lines joining the points of equal Percentage Depth Dose (PDD).