PHYSICAL PARAMETERS IN TELETHERAPY

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INTRODUCTION

- As the beam incident on a patient, the absorbed dose in the patient varies with depth

- Dose variation depends on many conditions such as beam energy, depth field size, SSD and beam collimation

- Calculation of dose in a patient involves considerations in regard to these parameters

PARAMETERS IN TELETHERAPY

- Dosimetric parameters used in the whole photon energy range:
  - Percentage depth dose (PDD)
  - Relative dose factor (RDF)
- Dosimetric parameters used at cobalt-60 and below:
  - Peak scatter factor (PSF)
  - Collimator scatter factor (Sc)
  - Phantom Scatter factor (Sp)
  - Scatter function (S)
  - Tissue air ratio (TAR)
  - Scatter air ratio (SAR)
- Dosimetric parameters used at cobalt-60 and above:
  - Tissue maximum ratio (TMR)
  - Tissue phantom ratio (TPR)
  - Scatter maximum ratio (SMR)

Percentage Depth Dose

- As a rule of thumb, an 18-MV, 6-MV, and ^{60}Co photon beam loose approximately 2%, 3.5% and 4.5% per cm

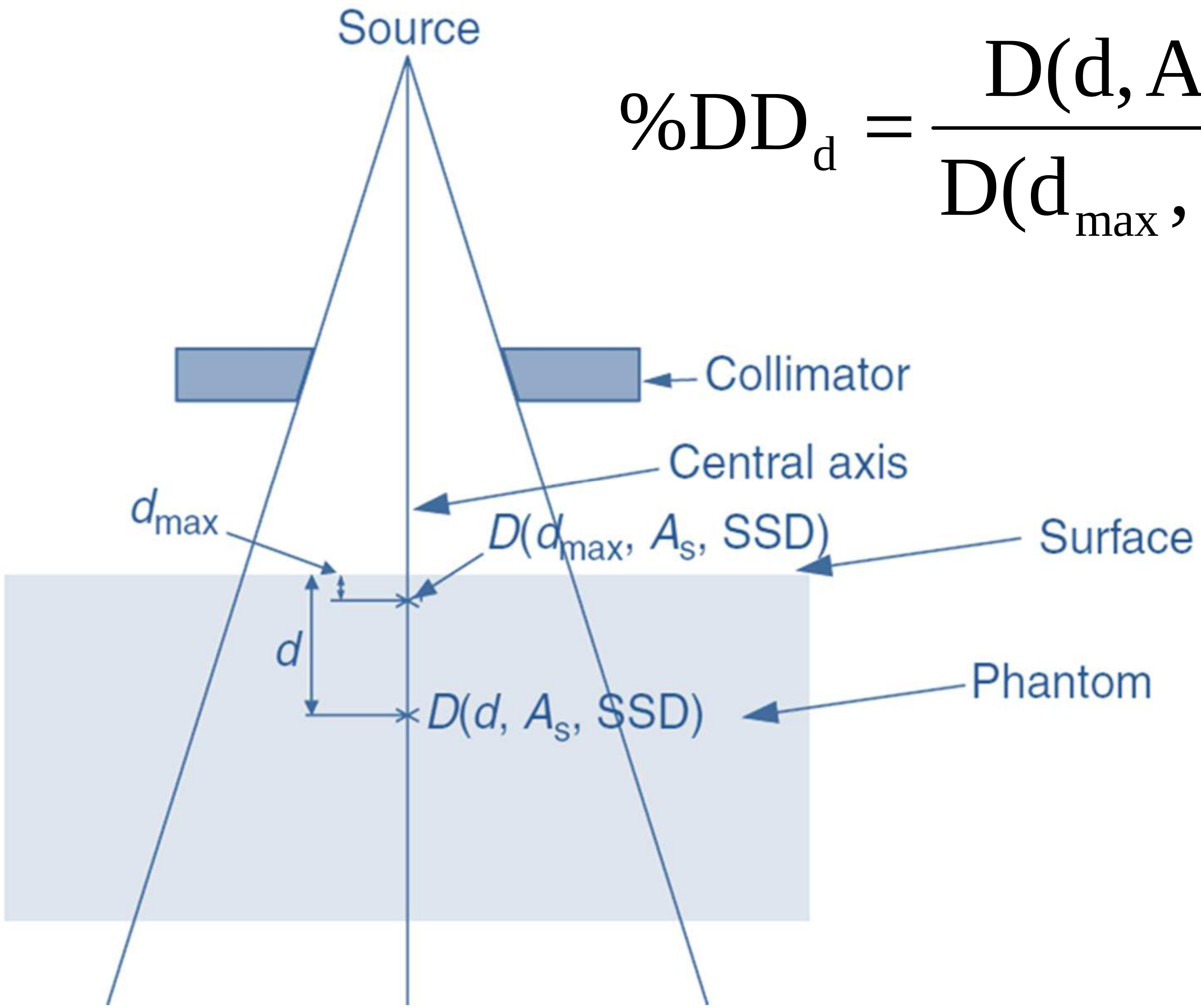
- PDD at any point is the dose at that point expressed as percentage of the peak dose on the central axis of the beam

- The ratio of the absorbed dose at a predefined depth (D_d) to absorbed dose maximum (D_{max}) in terms of % is known as PDD

$$\%DD_d = \frac{D(d)}{D(d_{max})} \times 100$$

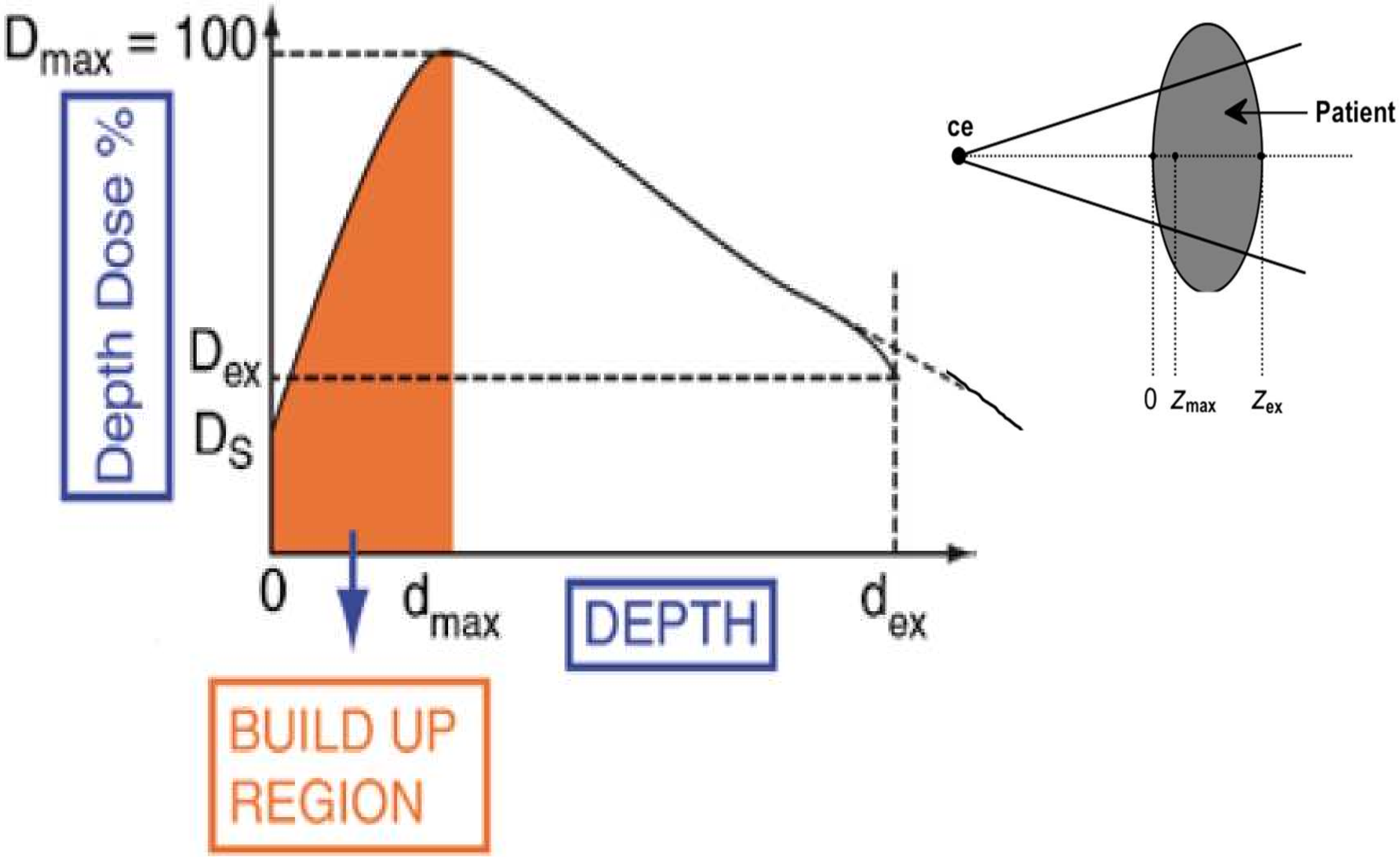
Percentage Depth Dose

$$\%DD_d = \frac{D(d, A_s, SSD)}{D(d_{max}, A_s, SSD)} \times 100$$



PDD Measurement (for Particular Field Size, SSD)

Percentage Depth Dose curve



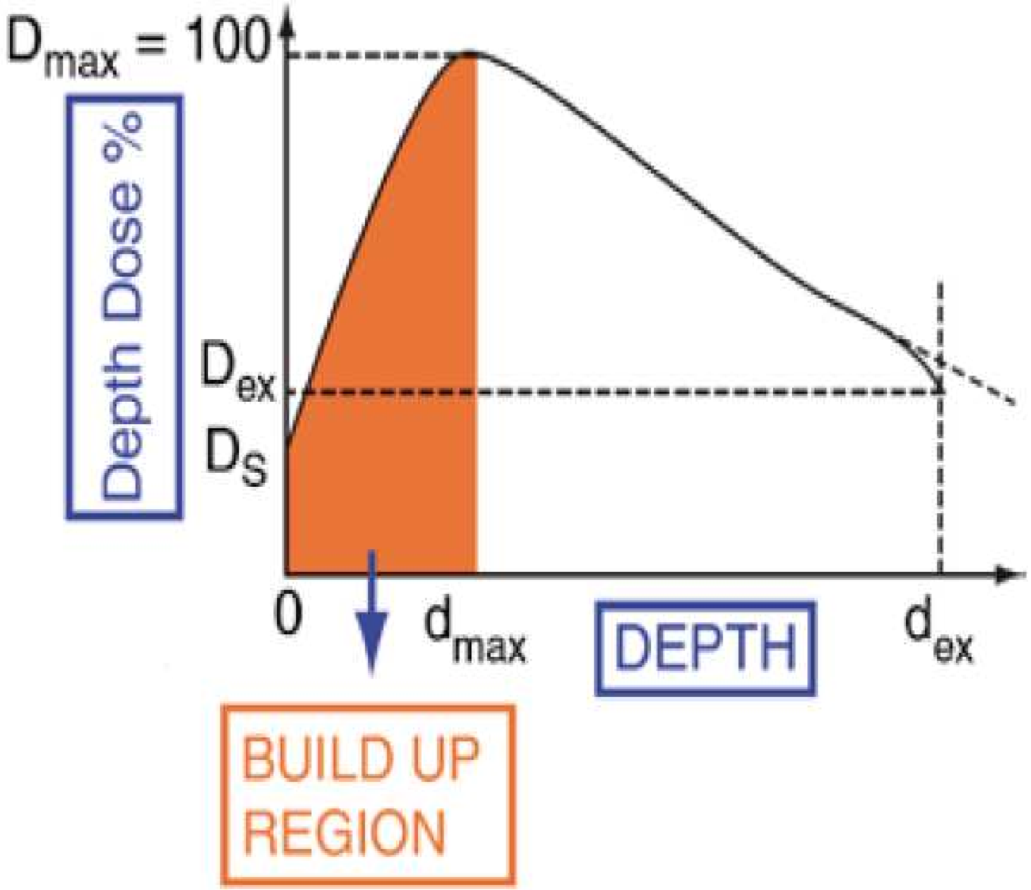
18MV PDD curve (10x10cm^2 field size, 100cm SSD)

Percentage Depth Dose – Surface Dose(D_S)

- Ortho-voltage and superficial beams deposit maximum dose on the skin surface
- For mega voltage photon beams, the surface dose is generally much lower than maximum dose which depends on energy and field size
- The larger the photon beam energy, lower the surface dose
- For a given beam energy, the surface dose increases with the field size
- The low surface dose compared with the maximum dose is referred to as the skin sparing effect which is advantage of megavoltage beams over Ortho-voltage beams

Percentage Depth Dose – **Build up Region**

- The dose region between the surface and depth d_{max} in MV photon beams is referred to as the dose build-up region
- This is due to relatively long range of energetic secondary charged particles that first released in the phantom by photon interactions



Percentage Depth Dose – **Depth of dose maximum (d_{max})**

- The depth of d_{max} depends on the beam energy and beam field size
- As the beam energy increases, the depth of d_{max} also increases
- Nominal values for d_{max} range from zero for superficial X-ray beams, through 0.5cm for Co-60 beam, 1.5cm for 6MV photon beam, 3.3cm for 18MV photon beam and 5cm for 25MV photon beam
- After d_{max} depth, the absorbed dose decreases due to attenuation and scatter

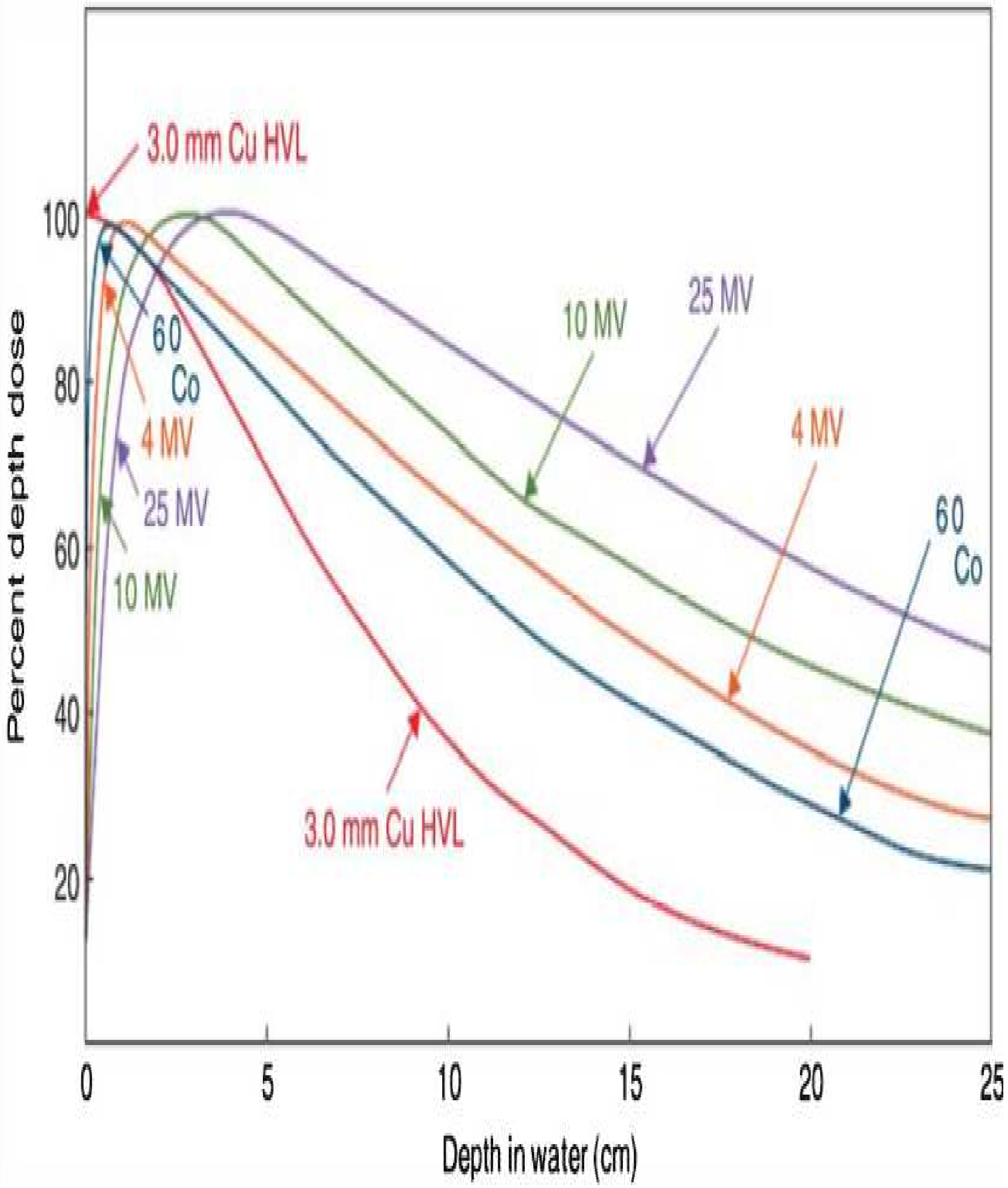
Energy	100 kV_p	350 kV_p	Co-60	4 MV	6 MV	10 MV	15 MV
z_{max}(cm)	0	0	0.5	1.0	1.5	2.7	2..3

Percentage Depth Dose - Dependency

- Various factors affect the central axis depth dose distribution in terms of PDD value, curve shape and d_{max} position etc.,

    - Beam Quality or Energy
    - Depth
    - Field Size and shape
    - Source to surface distance (SSD)
    - Beam Collimation

PDD- Dependence on Energy & Depth



- As beam energy increases, the penetration also increases which results in greater PDDs

ENERGY ↑ ➔ PDD↑

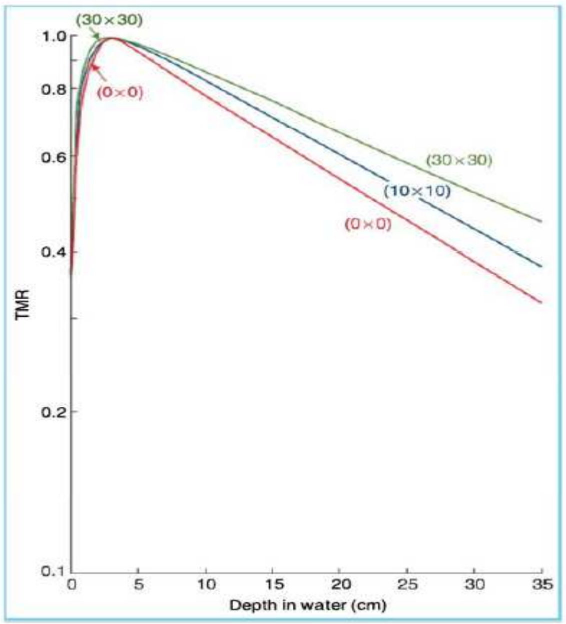
- As depth increases beyond the d_{max}, the attenuation, scattering and effect of inverse square law increases and PDD decreases

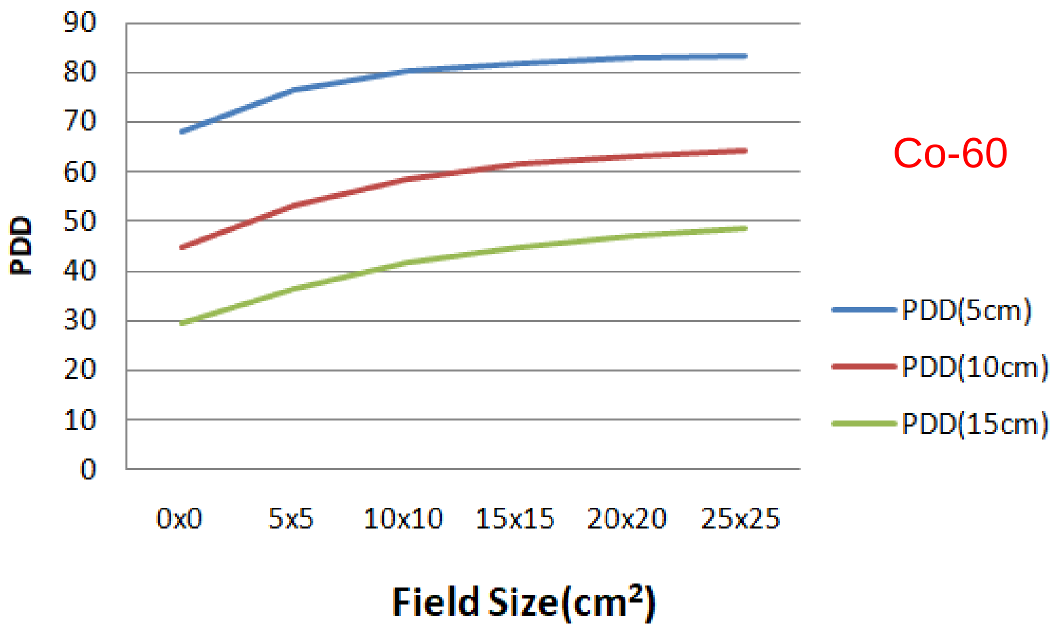
Depth↑ ➔ PDD ↓

PDD- Dependence on Field size and Beam Collimation

- For small fields, depth dose is caused by primary radiation only. But as the field size increases, the contribution of scattered photon to absorbed dose increases which causes increase in PDD

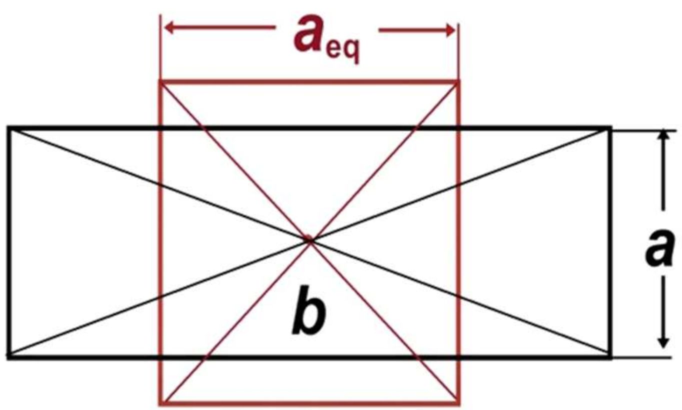
- The increase in PDD caused by increase in field size depends on beam quality Since scatter probability decreases with increase in energy, the field size dependence is less pronounced in high energy than lower energy beams.

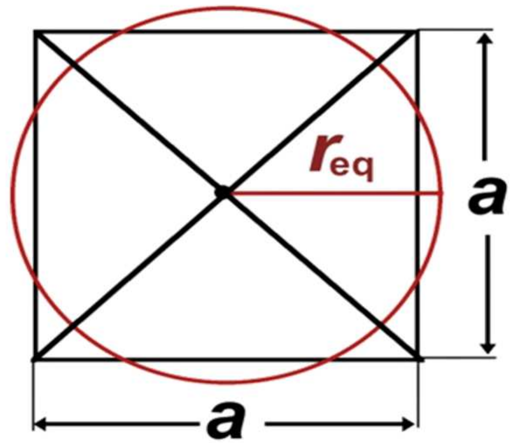




PDD- Dependence on Field size & Beam Collimation

- PDD data for radiation therapy beams usually tabulated for square fields
- But, the majority of the treatments encountered in clinical practice require rectangular or irregularly shaped fields
- A system of equating square fields to different field dimensions and shapes is required. So many empirical methods have been developed

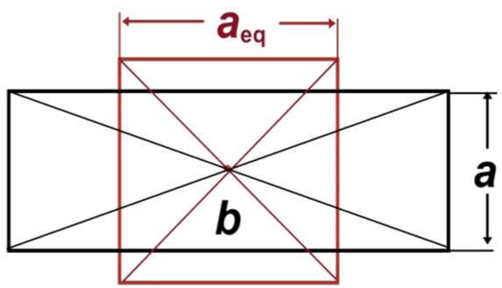




PDD- Dependence on Field size & Beam Collimation

- Equivalent square for rectangular field:
  - An arbitrary rectangular field with sides *a* and *b* will be approximately equal to a square field with side a_{eq} when both fields have the same area/perimeter ratio (Day's rule).

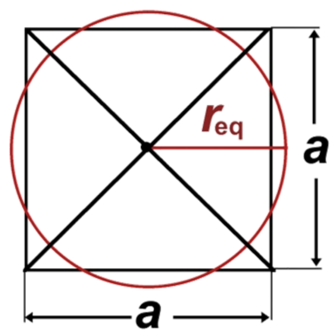
$$\frac{ab}{2(a+b)} = \frac{a_{eq}^2}{4a_{eq}}$$



$$a_{eq} = \frac{2ab}{a+b}$$

- Equivalent circle for square field:
  - An arbitrary square field with side *a* will be equivalent to a circular field with radius r_{eq} when both fields have the same area.

$$a_{eq}^2 = \pi r_{eq}^2$$



$a = \sqrt{\pi}.\ r_{eq}$

PDD- Dependence on SSD

- PDD increases with SSD as a result of the inverse square law.

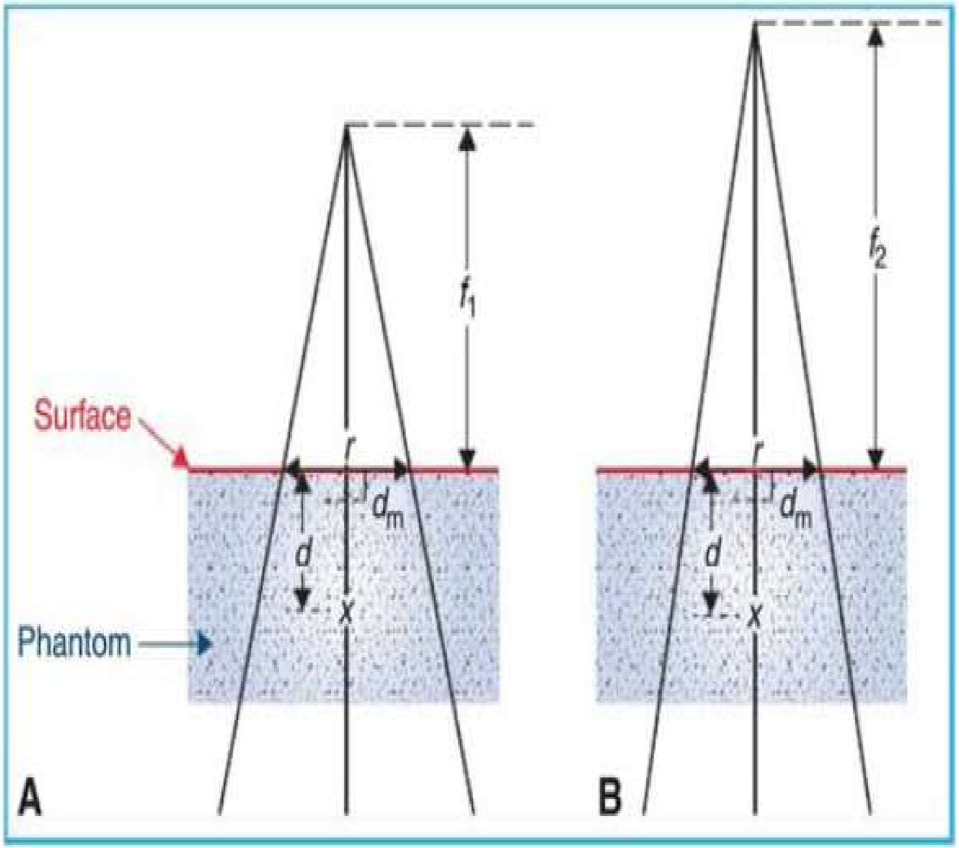
- Although the actual dose rate at a point decreases with increase in distance from the source, the PDD, which is a relative dose with respect to a reference point, increases with SSD

SSD (cm)	60	80	100	120	140
PDD (5cm, 10x10cm^2)	76.2	78.8	80.0	81.3	82.3

PDDs for Co-60 beam in water for various SSDs

PDD- Non-Standard SSD

- PDDs for a standard SSD may needs to convert to various SSDs for different clinical situations.
- Mayneord F factor is used to convert PDD of one SSD to other SSD which accounts inverse square law only , without considering changes in scattering



$$P(d,r,f_1) = 100.\frac{(f_1+d_m)^2}{(f_1+d)^2}.c^{-\mu d}.K_s$$

$$P(d,r,f_2) = 100.\frac{(f_2+d_m)^2}{(f_2+d)^2}.c^{-\mu d}.K_s$$

dividing the above two equations,

$$\frac{P(d,r,f_2)}{P(d,r,f_1)} = \frac{(f_2+d_m)^2}{(f_1+d_m)^2}.\frac{(f_1+d)^2}{(f_2+d)^2} = F$$

$F > 1$ in the case of $f_2 > f_1$

$F < 1$ in the case of $f_2 < f_1$

PDD - Summary

- PDD increases with beam energy
    - Due to high penetration
- PDD increases with Field size
    - Due to increase in scatter component
- PDD increases with SSD
    - Due to less effectiveness of inverse square law
- PDD decreases with depth(beyond d_{max})
    - Due to exponential attenuation

- Due to PDD greatly depends on SSD, PDD can not be used in SAD setup
- This causes introduction of other parameters which does not depends on SSD

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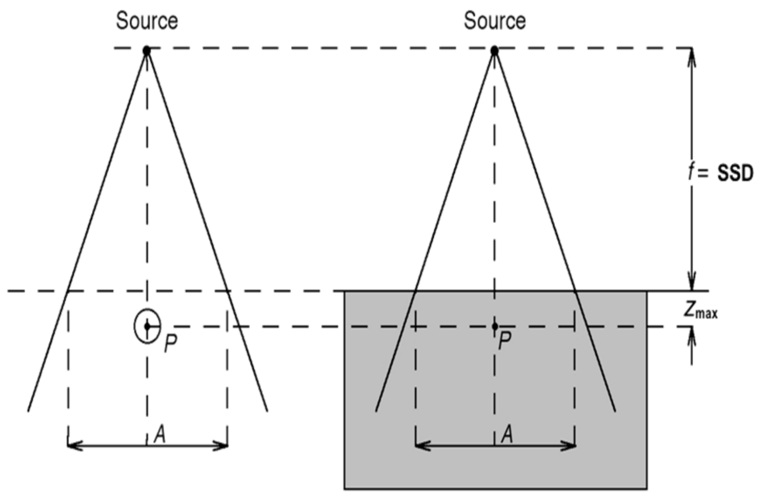
Peak Scatter Factor (PSF)

- PSF is defined as the ratio of dose to a point at Dmax in phantom to that of dose in air at the same point

$$\mathrm{PSF}(A,h\nu)=\frac{D_{\mathrm{P}}(z_{\max},A,f,h\nu)}{D'_{\mathrm{P}}(A,h\nu)}$$

PSF gives the factor by which the radiation dose at point P in air is increased by scattered radiation when point P is in the phantom at depth $z_{\max}$.



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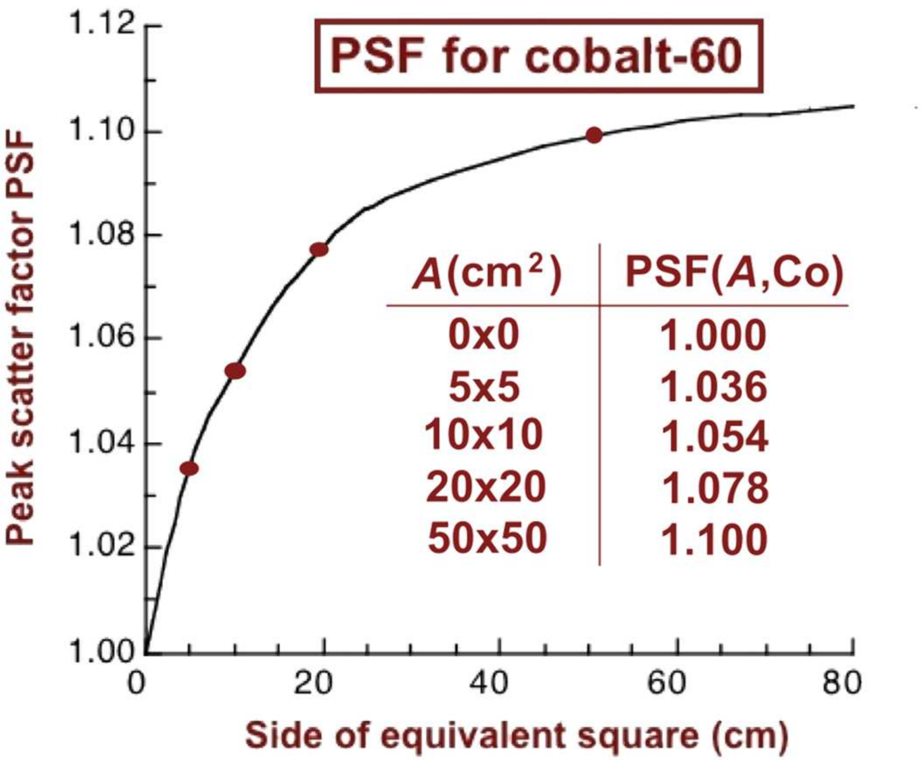
Back Scatter Factor (BSF)

- At low photon energies, z_{max} is on the phantom surface (z_{max} = 0) and the PSF is referred to as the backscatter factor BSF.

- PSF for field size of zero area is equal to 1 for all photon beam energies, i.e.,

- As the field size increases, PSF first increases from unity linearly as field size increases and then saturates at very large fields.



Scattered radiation Dose

- The dose to a point in a medium may be analyzed into primary and scattered components
- The primary dose is contributed by the original photons emitted from the source
- The scattered dose is the result of collimator and phantom components
- The scattered dose results from collimator is represented by a factor Collimator Scatter Factor (S_c) where as scattered dose originated from phantom is quantified by Phantom Scattered Factor (S_p)

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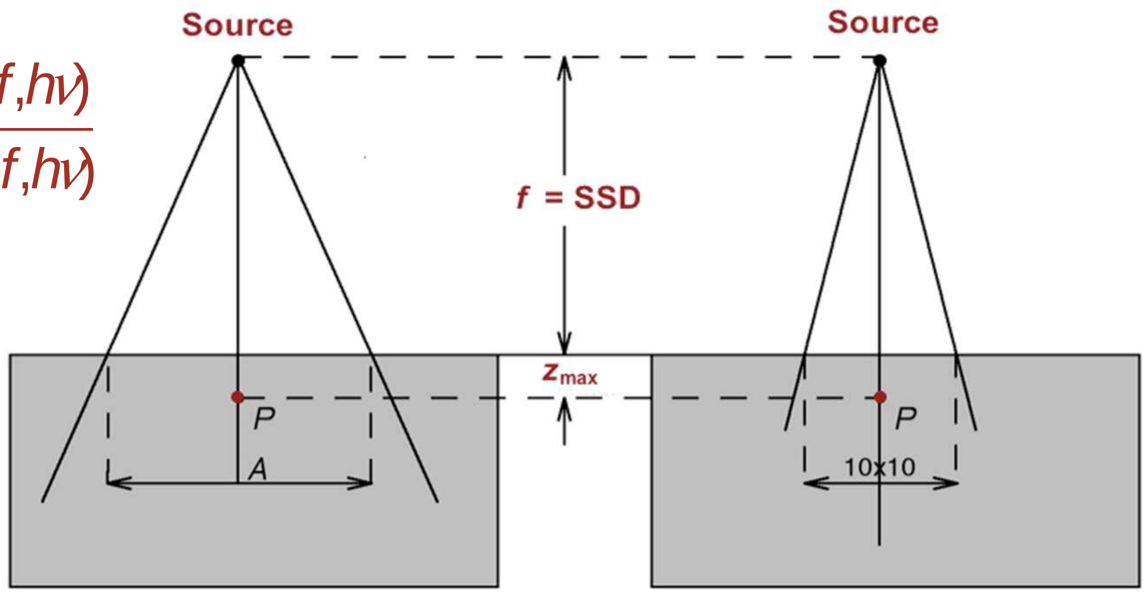
Relative Dose Factor (Sc,p)

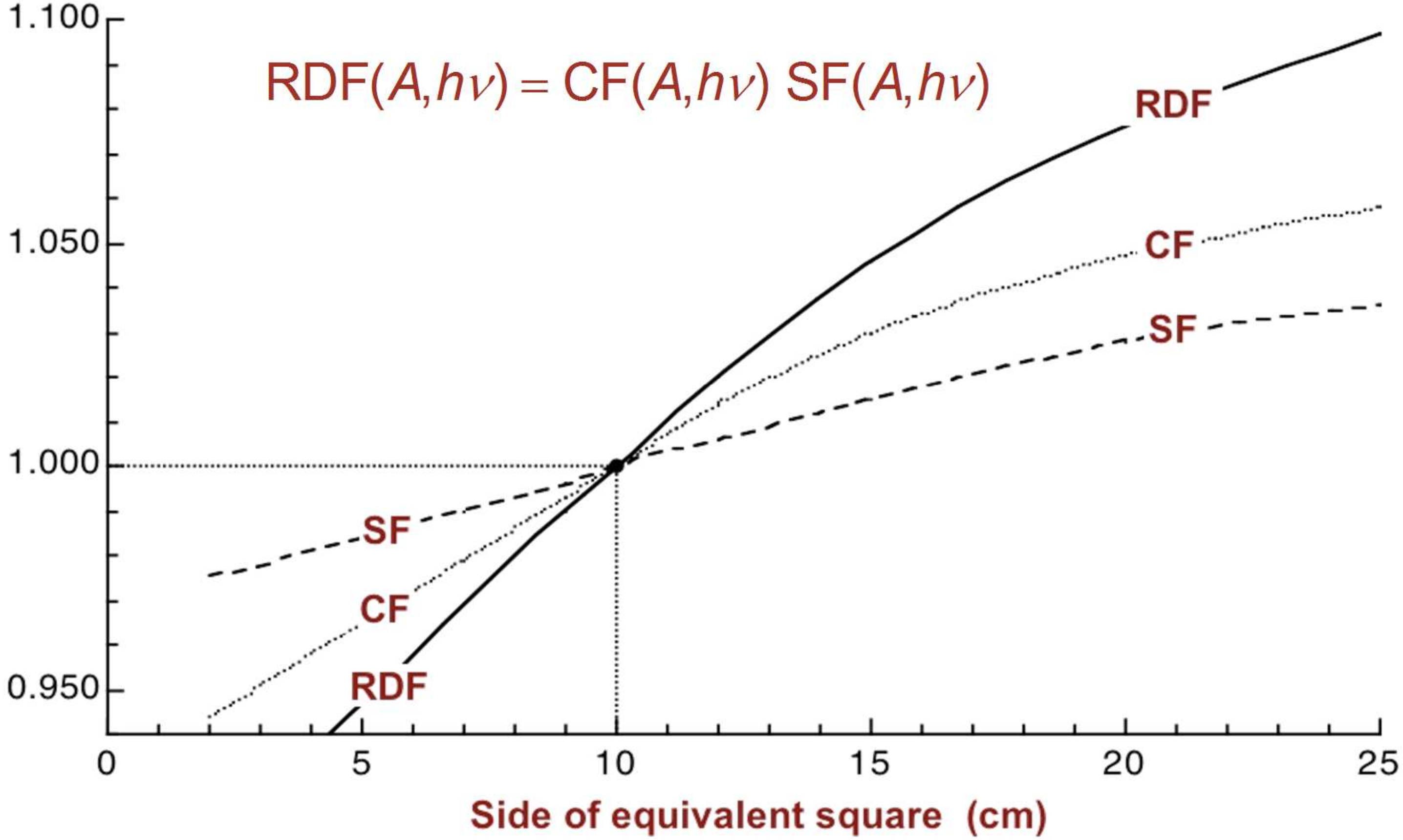
- The ratio of the dose at point P for field size *A* to the dose at point P for field size 10x10 cm^2 is called the relative dose factor RDF or total scatter factor $S_{c,p}$ in Khan's notation or machine output factor OF

$$RDF(A,h\nu) = S_{c,p}(A,h\nu) = \frac{D_P(z_{max},A,f,h\nu)}{D_P(z_{max},10,f,h\nu)}$$



Relative Dose Factor (Sc,p)

- Typical values for Sc,p , Sc and Sp for a cobalt-60 gamma ray beam



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Collimator Scatter Factor (S_c) or CF

- The ratio of the dose measured in any field at D_{max} in air to D_{max} measured in a reference field ($10x10cm^2$) in air is called Collimator Scatter factor (S_c)

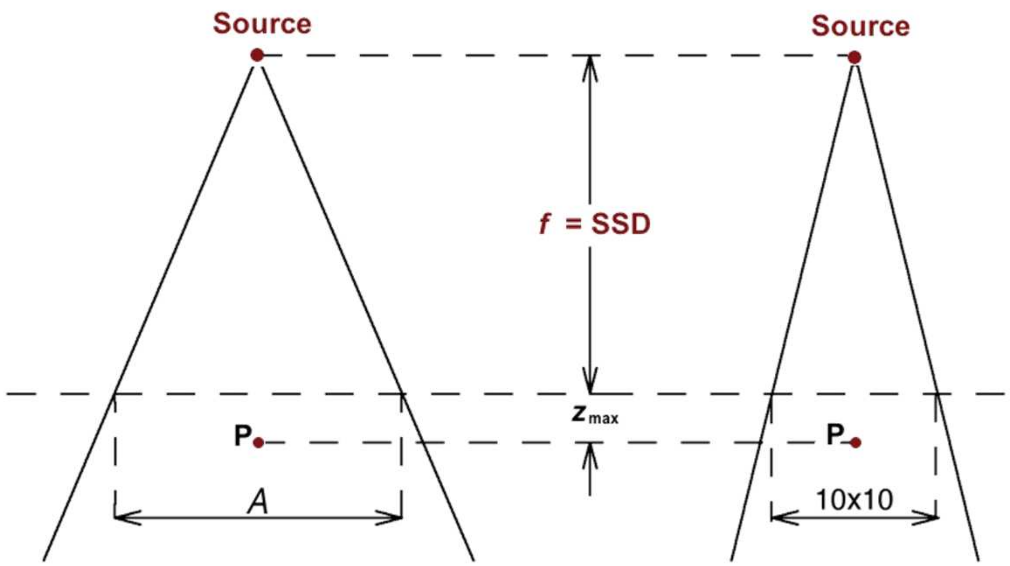
- S_c measured with Ion chamber with a build up cap of a size large enough to provide maximum dose build up for a given beam

CF is normalized to 1 for the nominal field of 10x10 cm^2 at the nominal SSD for the treatment machine.

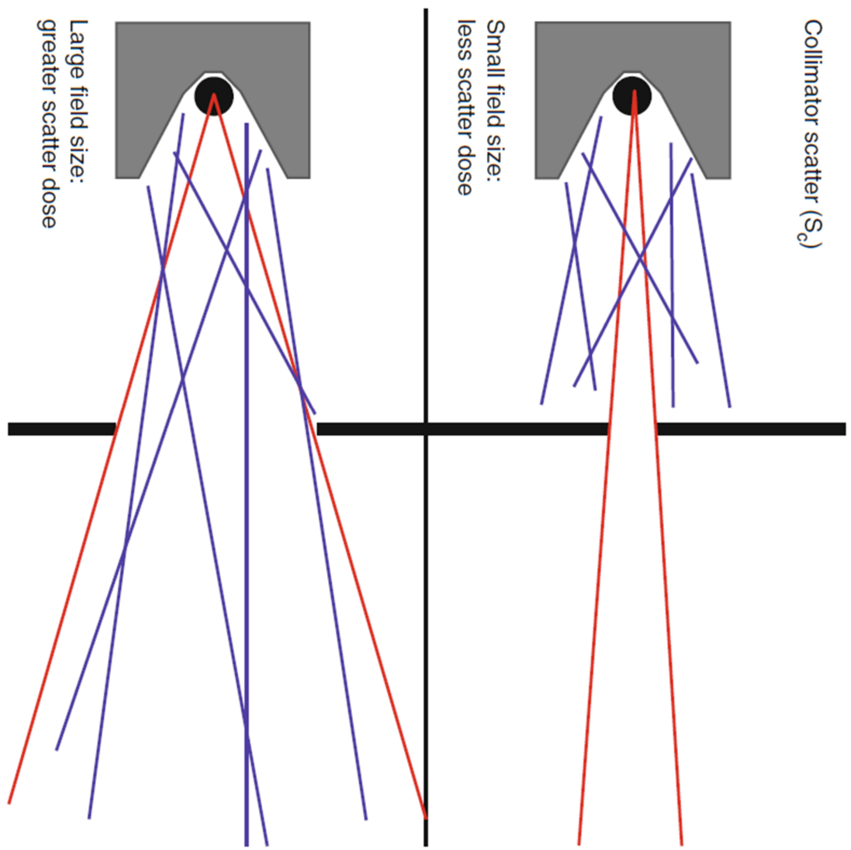
CF > 1 for fields *A* exceeding 10x10 cm^2.

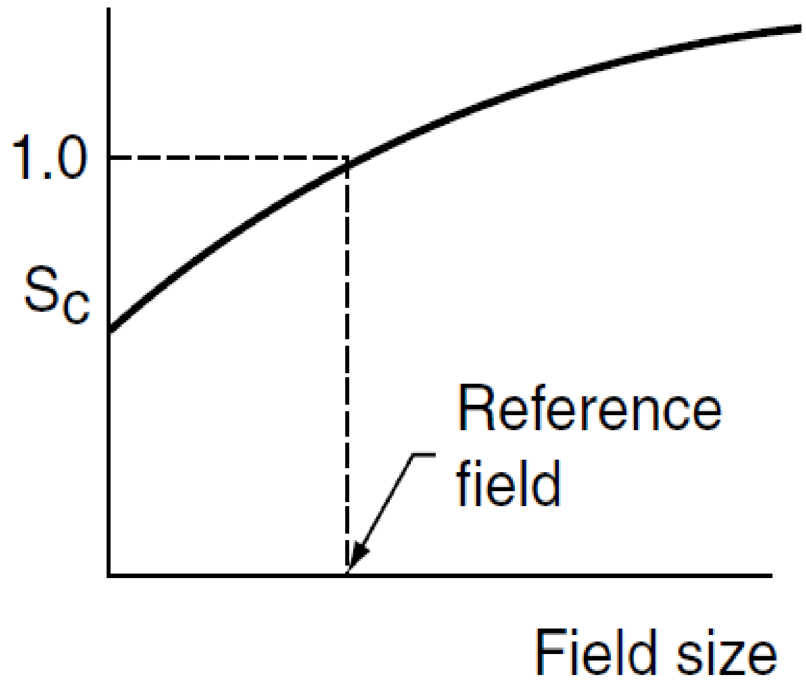
CF = 1 for 10x10 cm^2 field.

CF < 1 for fields *A* smaller than 10x10 cm^2.



Collimator Scatter Factor (S_c)





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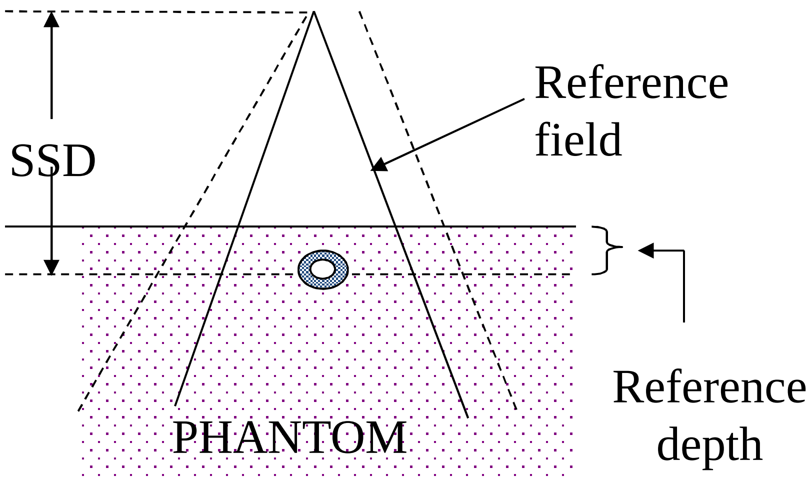
Phantom Scatter Factor (Sp)

- The change in scatter radiation originating in the phantom reference depth as the field side is change
- Definition
  - The ratio of the dose for a given field at a reference depth (e.g. depth of D_{max}) to the dose rate at the same depth of the reference field size (10 x 10 cm), with the same collimator opening in phantom
- Related to the change in the volume of the phantom irradiated



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Scatter Function (s)

- The scatter component at point Q in the phantom is determined as follows:

$$\text{Scatter component at Q} = \text{Total dose at Q} - \text{Primary dose at Q}$$

- The scatter function $S(z,A,f,h\nu)$ is defined as the scatter component at point Q normalized to 100 cGy of primary dose at point P (Z max)

$$S(z,A,f,h\nu) = \frac{\text{Scatter component at Q}}{D'_{P}(=100\ \text{cGy})}:$$

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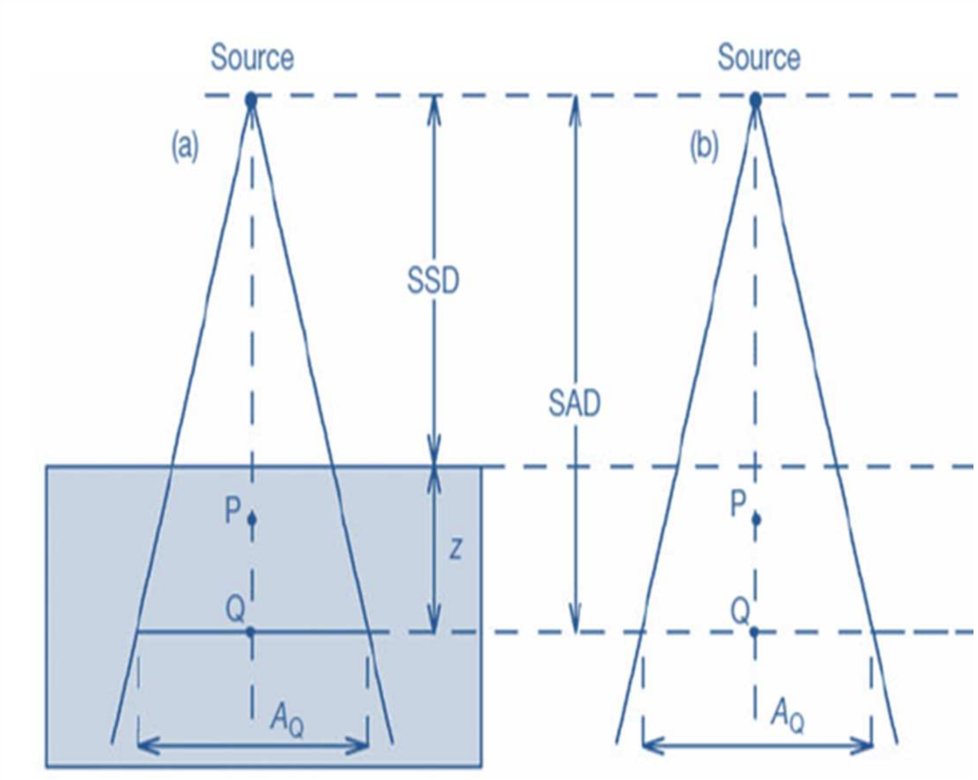
Tissue Air Ratio (TAR)

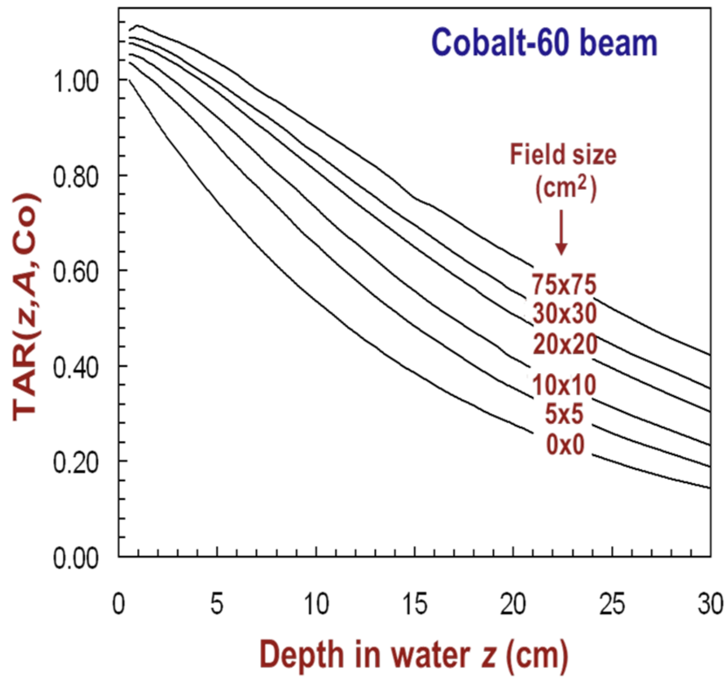
- Tissue Air Ratio (TAR) defined to remove the SSD dependence
- TAR may be defined as the ratio of the dose (D) at a given point in the phantom to the dose in free space (D_{fs}) at the same point measured with build up cap

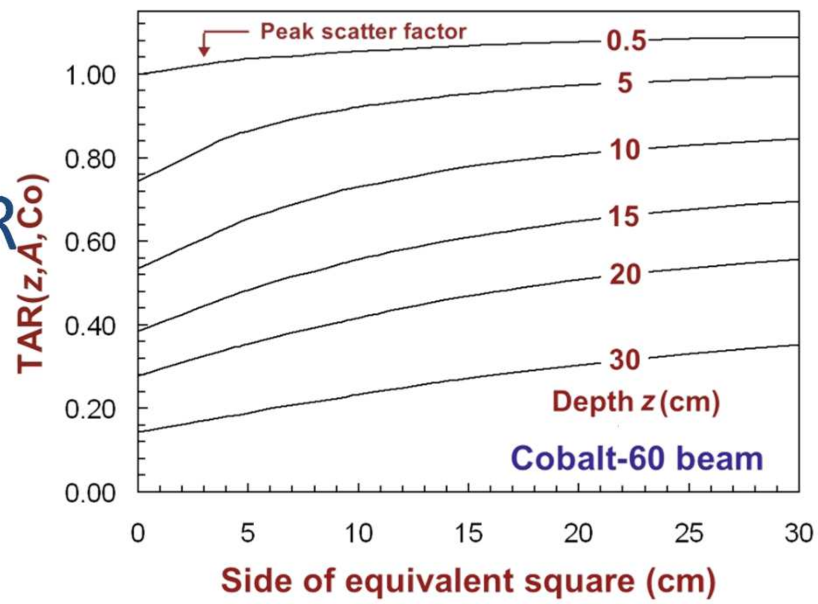


$$TAR(Q) = \frac{D_{Q,\,phantom}}{D_{Q,\,freespace}}$$

TAR

- As like PDD, TAR also depends on various parameters except SSD
  - As depth increases, TAR decreases
  - As field size increases, TAR increases
  - As energy increases, TAR increases
  - No change in TAR while SSD varies





TAR

Drawbacks of TAR

- The Concept of TAR is not recommended for use with MV beams above ^{60}Co and 4 MV because of difficulty in air measurements with large size build-up cap

- Usually, the buildup cap is different from the phantom material which induces uncertainty in the TAR measurements

- The measurement of TARs are cumbersome due to water & air measurements

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Scatter Air Ratio (SAR)

- TAR(z,A_Q, hν) consists of two components:
  - Primary component TAR(z,0, hν) for zero field size
  - Scatter component referred to as scatter-air ratio SAR(z,A_Q, hν)

$$\mathrm{SAR}(z, A_Q, h\nu) = \mathrm{TAR}(z, A_Q, h\nu) - \mathrm{TAR}(z, 0, h\nu)$$

- The SAR gives the scatter contribution to the dose at point Q in a water phantom per 1 cGy of dose to a small mass of water at point Q in air.

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Tissue-Phantom Ratio (TPR)

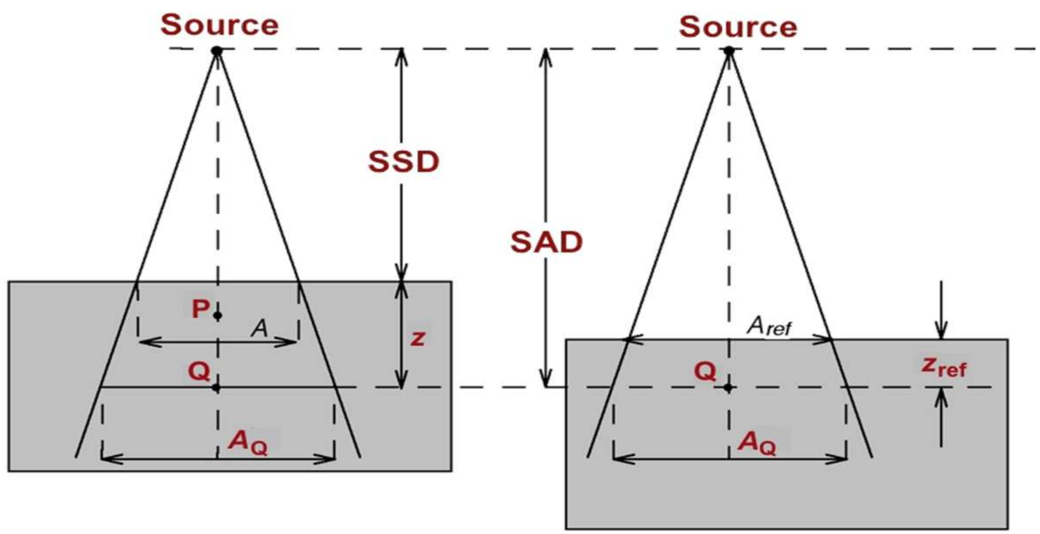
- For isocentric setups with megavoltage photon above 4 MV energies the concept of tissue-phantom ratio TPR was developed

- Similarly to TAR the TPR depends upon z, A_Q, and $h\nu$

$$\mathrm{TPR}(z, A_Q, h\nu) = \frac{D_Q}{D_{Q_{ref}}}$$

  - D_Q is the dose at point Q at depth z

  - D_{Qref} is the dose at depth z_{ref}.



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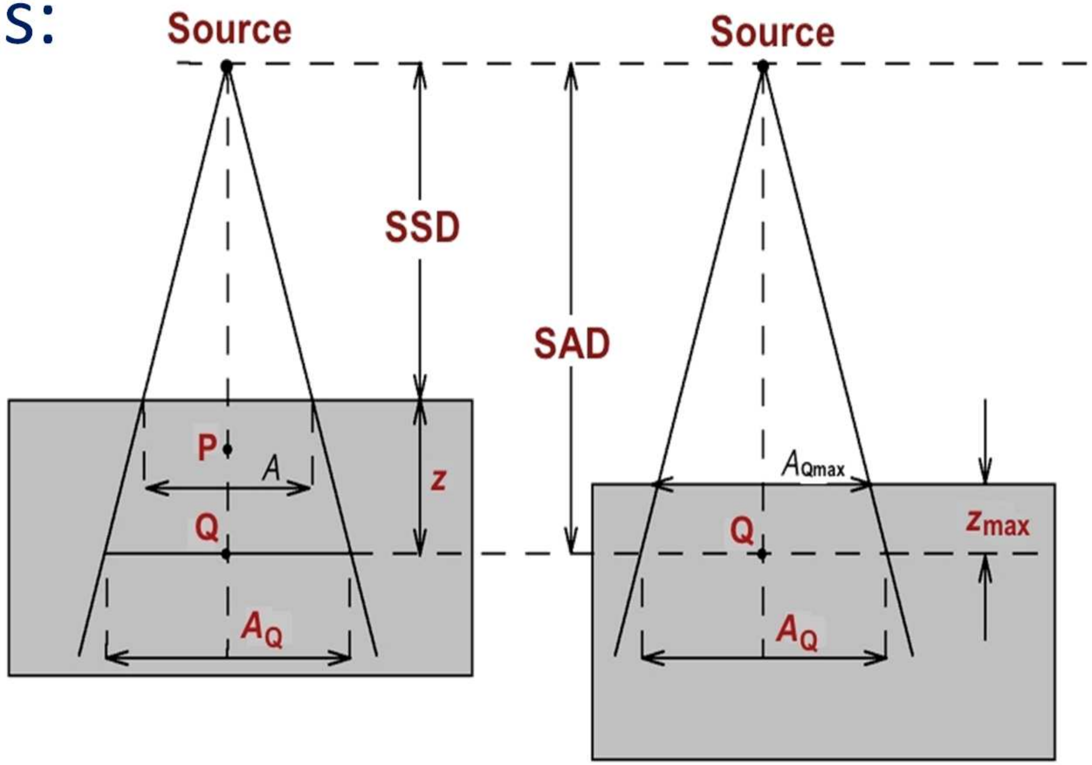
Tissue-Maximum ratio (TMR)

- Tissue-maximum ratio TMR is a special TPR for $z_{ref} = z_{max}$.
- TMR is defined as:

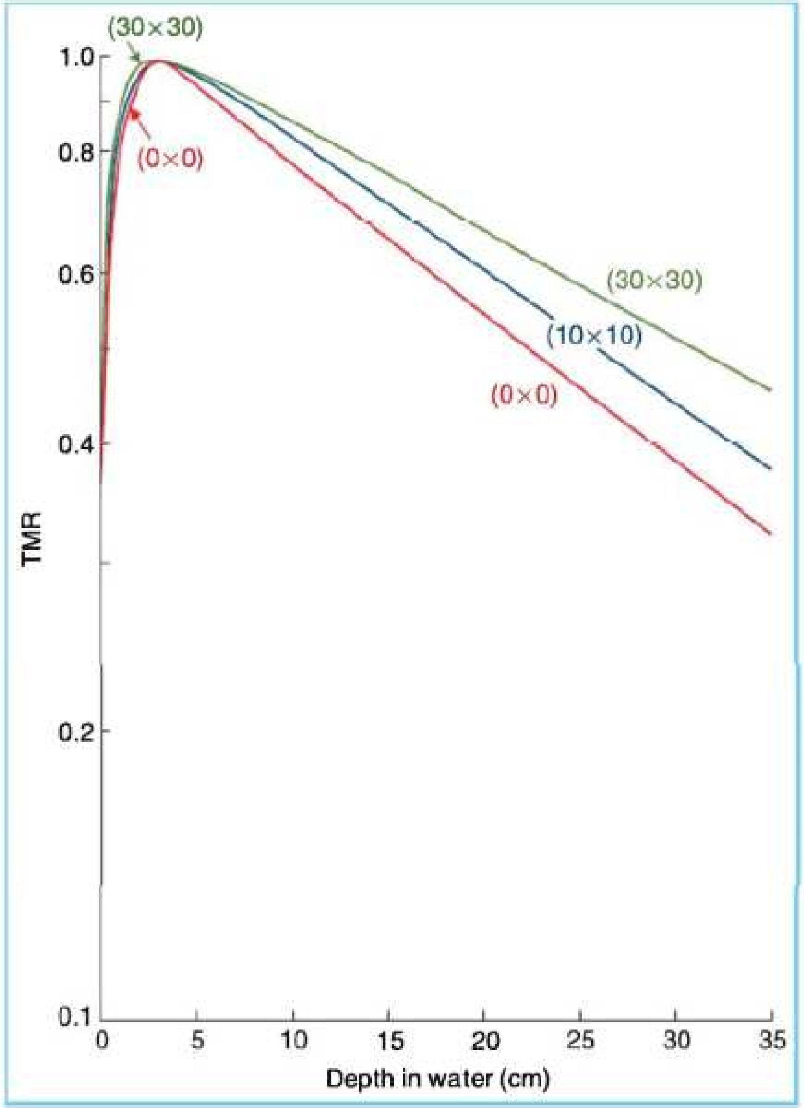
$$\mathrm{TMR}(z, A_Q, h\nu) = \frac{D_Q}{D_{Q_{max}}}$$

  - D_Q is the dose at point Q at depth z
  - D_{Qmax} is the dose at depth z_{max}.



TPR & TMR

- As like PDD, TPR/TMR also depends on Beam energy, field size and depth but not on SSD
- As Beam energy increases, TMR increases
- As Field size increases, TMR increases
- As depth increases, TMR decreases
- No change in TMR as SSD changes



PARAMETERS IN TELETHERAPY

- Dosimetric parameters used in the whole photon energy range:
  - Percentage depth dose (PDD)
  - Relative dose factor (RDF)
- Dosimetric parameters used at cobalt-60 and below:
  - Peak scatter factor (PSF)
  - Back scatter factor (BSF)
  - Collimator scatter factor (Sc)
  - Phantom Scatter factor (Sp)
  - Scatter function (S)
  - Tissue air ratio (TAR)
  - Scatter air ratio (SAR)
- Dosimetric parameters used at cobalt-60 and above:
  - Tissue phantom ratio (TPR)
  - Tissue maximum ratio (TMR)
  - Scatter maximum ratio (SMR)

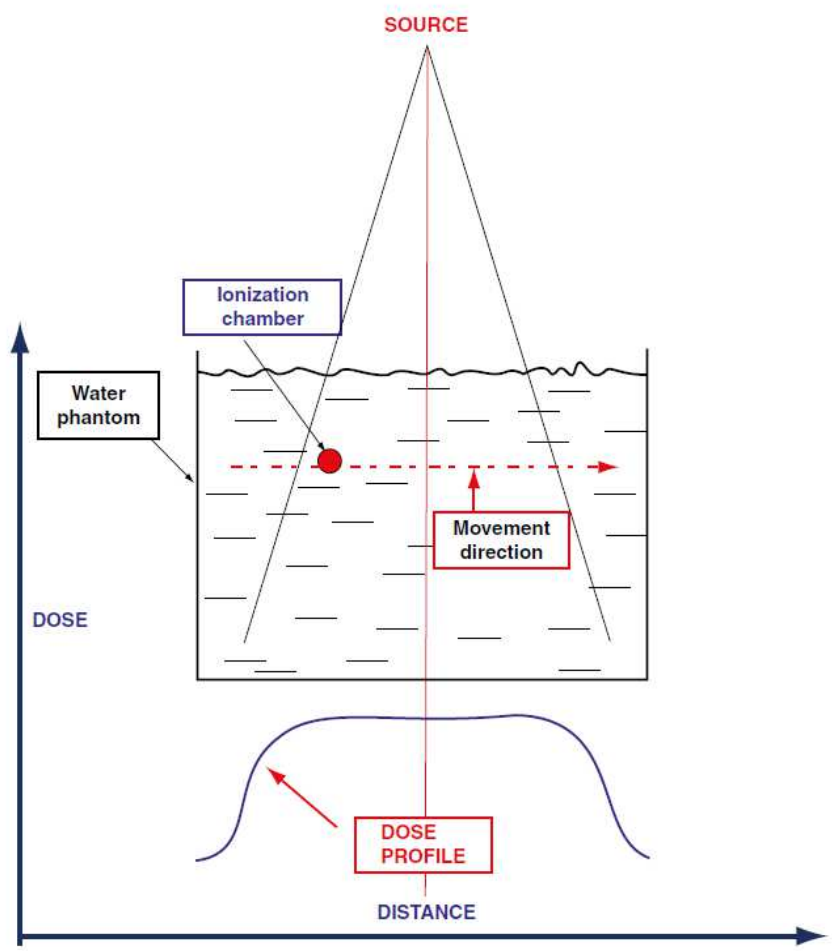
Scatter Maximum Ratio (SMR)

- TMR(z,A_Q, hν) can be separated into the primary component TMR(z,0, hν) and the scatter component called the scatter-maximum ratio SMR(z,A_Q, hν).

SMR (Z,AQ, hv) = TMR (Z,AQ, hv) - TMR (Z,0, hv)

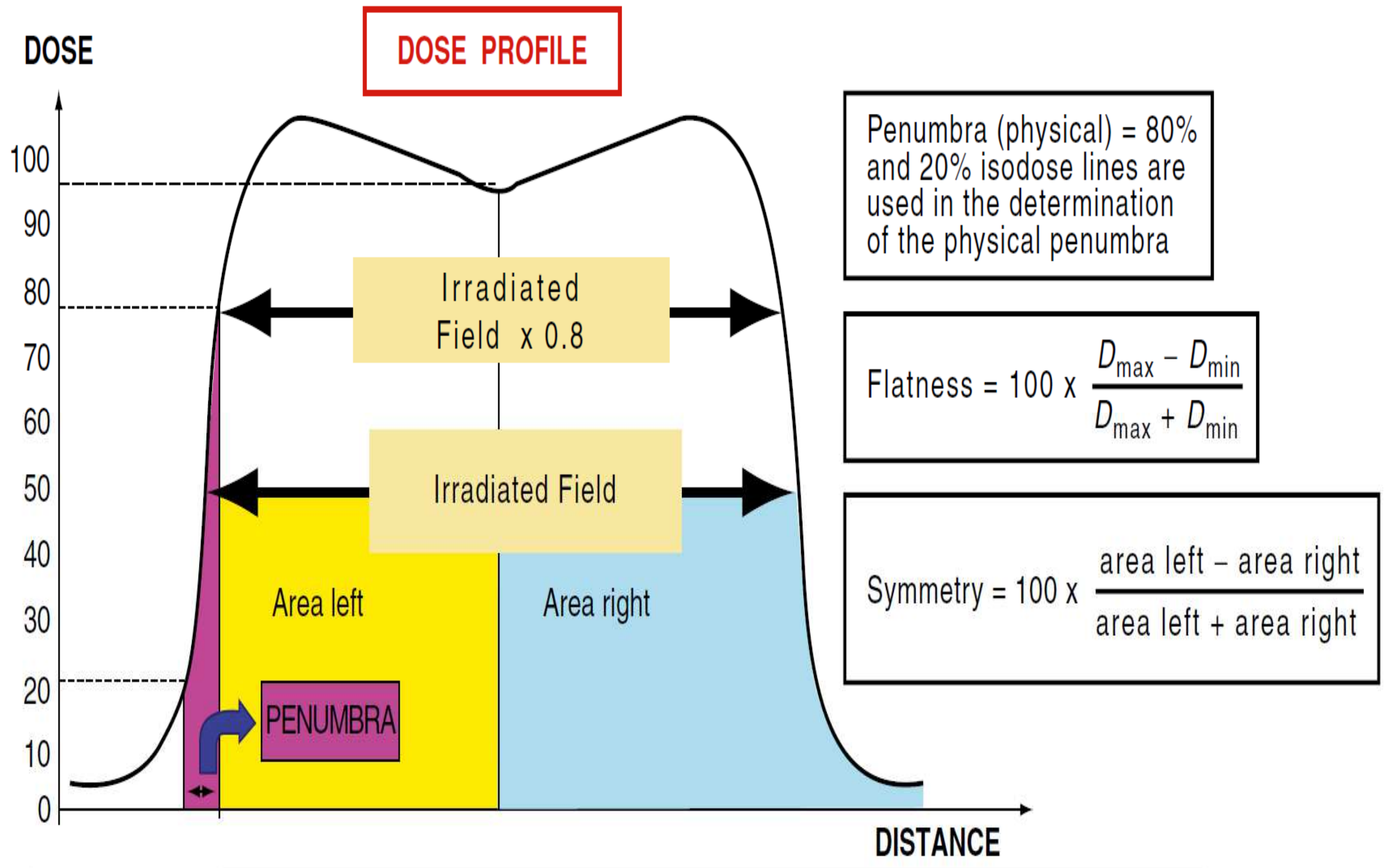
Off Axis Profiles

- Dose distribution along the beam central axis give only part of the information required for an accurate dose description inside the patient.
- Dose distributions in 2D and 3D are determined with central axis data in conjunction with off-axis dose profiles.
- Off axis profiles measured perpendicularly to the beam central axis at a given depth in a phantom



Off Axis Profiles

The Off Axis profiles are characterized by various parameters such as Flatness, Symmetry, and Penumbra etc.,



Off Axis Profiles: Flatness

- The beam flatness is assessed by finding the maximum D_{max}, and minimum D_{min} dose point values on the beam profile within the central 80% of the beam width and then,

$$F = 100 \times \frac{D_{\max} - D_{\min}}{D_{\max} - D_{\min}}$$

- Generally flatness should be less than 3% when measured in a water phantom at a depth of 10cm and an SSD of 100cm for the largest field size (40x40 cm^2)

Off Axis Profiles: Symmetry

- Beam is symmetrical when any points on a beam profile, equidistant from the central axis point, have same dose value

or

- The areas under the d_{max} beam profile on each side (left & right) of the central axis extending to the 50% dose level are determined and then

$$S = 100 \times \frac{area_{left} - area_{right}}{area_{left} + area_{right}}$$

- The acceptable symmetry value is around 2%
- Beam symmetry is usually determined at d_{max}.

Off Axis Profiles: Penumbra

- The penumbra is defined as the region of steep dose rate decrease at the edge of radiation beam

- Depending on various factors, the penumbra is divided into three types such:

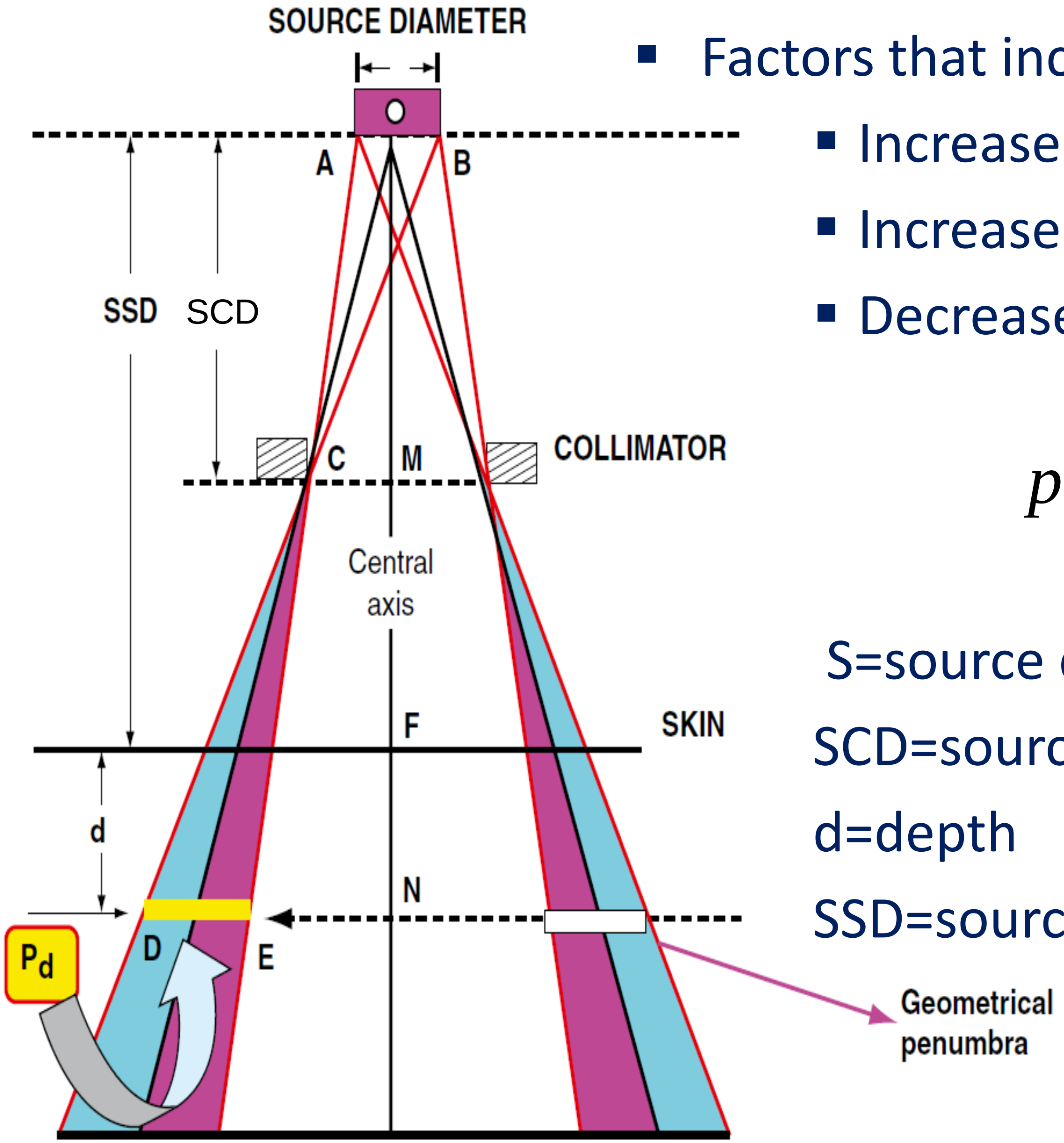
  - Geometrical Penumbra : This is due to finite size of the source or focal spot size

  - Transmission Penumbra: This occurs due to partial transmission of beam through collimators

  - Scatter Penumbra: This is due to significant scatter arising from the phantom

- The total penumbra is referred to as the Physical penumbra and the sum of the three individual penumbras

Off Axis Profiles: Penumbra



- Factors that increase the penumbra:
  - Increase in SSD
  - Increase in source size
  - Decrease in SCD

$$p = \frac{s(SSD + d - SCD)}{SCD}$$

S=source diameter

SCD=source to collimator distance

d=depth

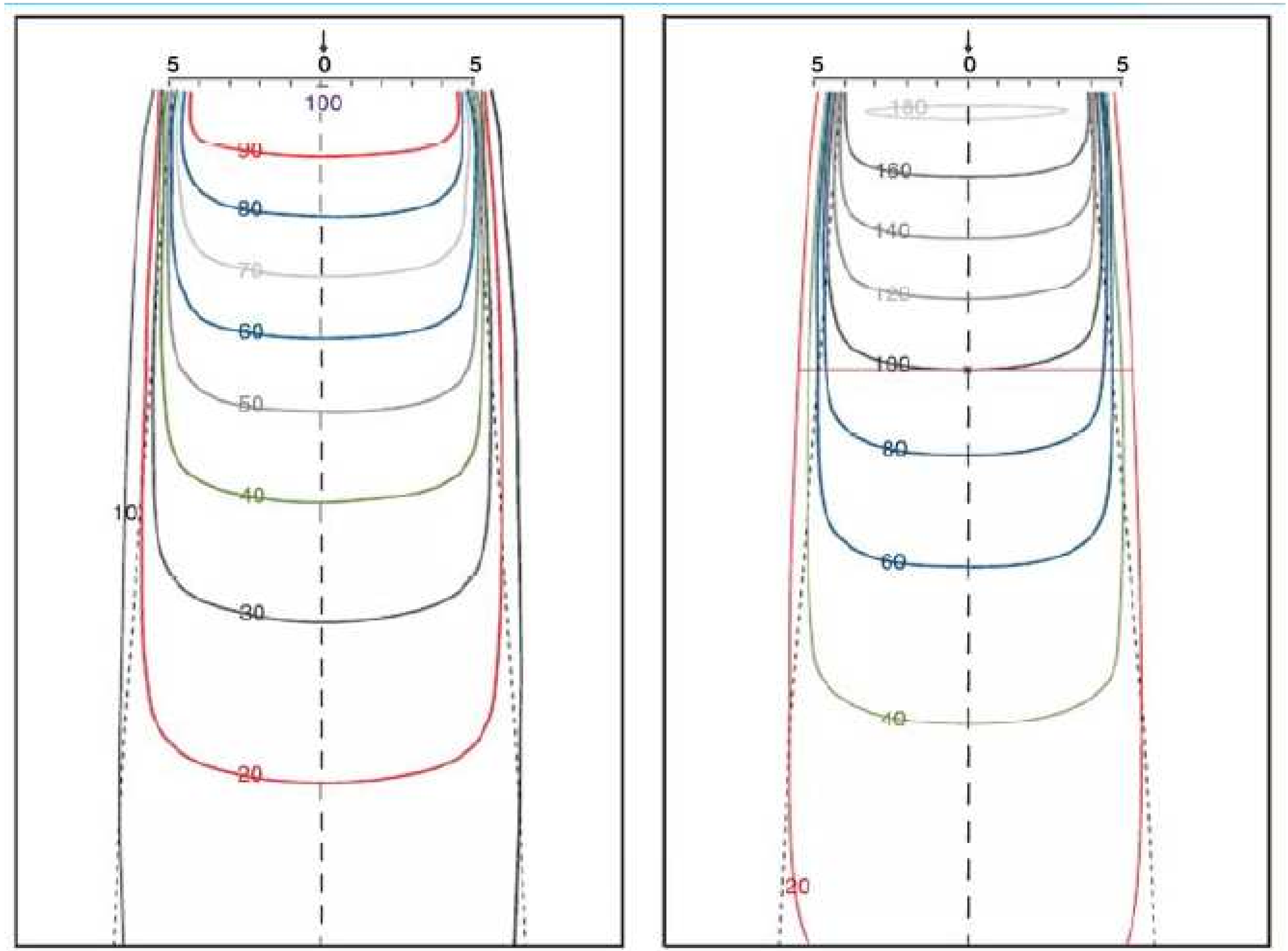
SSD=source to surface distance

Isodose curves

- Volumetric or planar variations in absorbed dose distributions are depicted by means of Isodose curves
- Isodose lines are lines passing through points of equal dose, usually drawn at regular intervals of absorbed dose and expressed as a percentage of the dose at a reference point
- An Isodose chart for a given single beam consists of a family of isodose curves usually drawn at regular increments of PDD

Isodose curves

- For SSD setups, all isodose values are normalized to 100% dose point on the central beam axis at d_{max}
- For SAD setups, the isodose values are normalized to isocenter

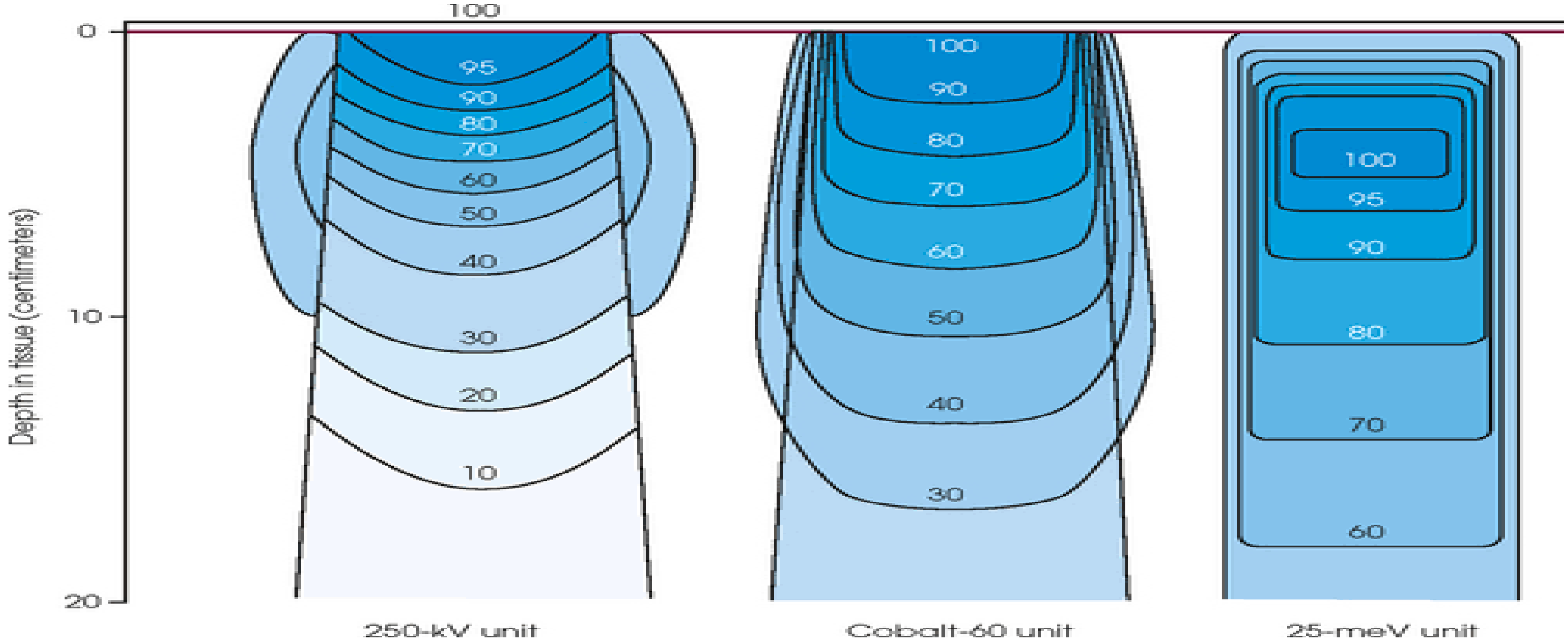


Isodose curves

- The dose at any depth is greatest on the central axis of the beam and gradually decreases towards edges of the beam
- Near the edges of the beam, the dose rate decreases rapidly as a function of lateral distance from the beam axis. This is due to geometric penumbra and by reduced side scatter
- Outside the geometric limits of the beam, the dose variation is the result of side scatter from the field and both leakage and scatter from the collimator

Isodose curves

- The shape of isodose curves is greatly depends on
  - Beam energy
  - Source size
  - Beam collimation
  - Field Size
  - SSD
  - Source to Collimator distance



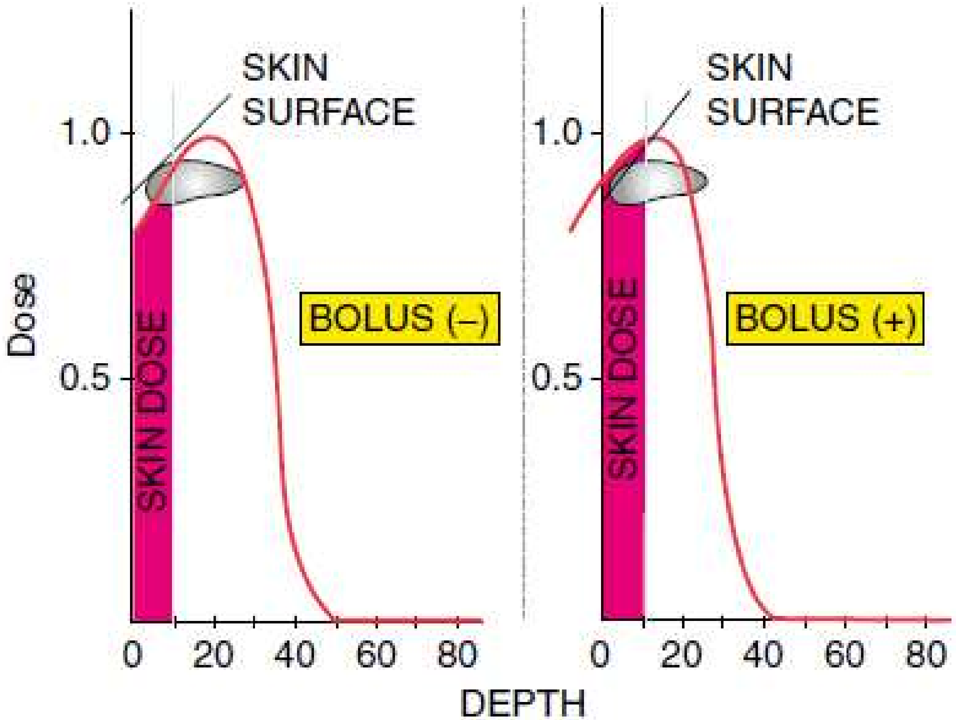
Corrections for irregular contours

- A radiation beam striking an irregular patient surface produces an isodose distribution that differs from the standard distribution obtained on flat surfaces (i.e. Dose distribution is not homogeneous if the surface of the patient is not flat)
- Two approaches are used to address this problem
- The irregular body surface may be compensated by introducing various materials
  - Bolus
  - Compensators
  - Wedge filters
- The effect can be corrected through various calculation methods
  - Effective SSD method
  - TAR or TMR method
  - Isodose shift method

Corrections for irregular contours: Bolus

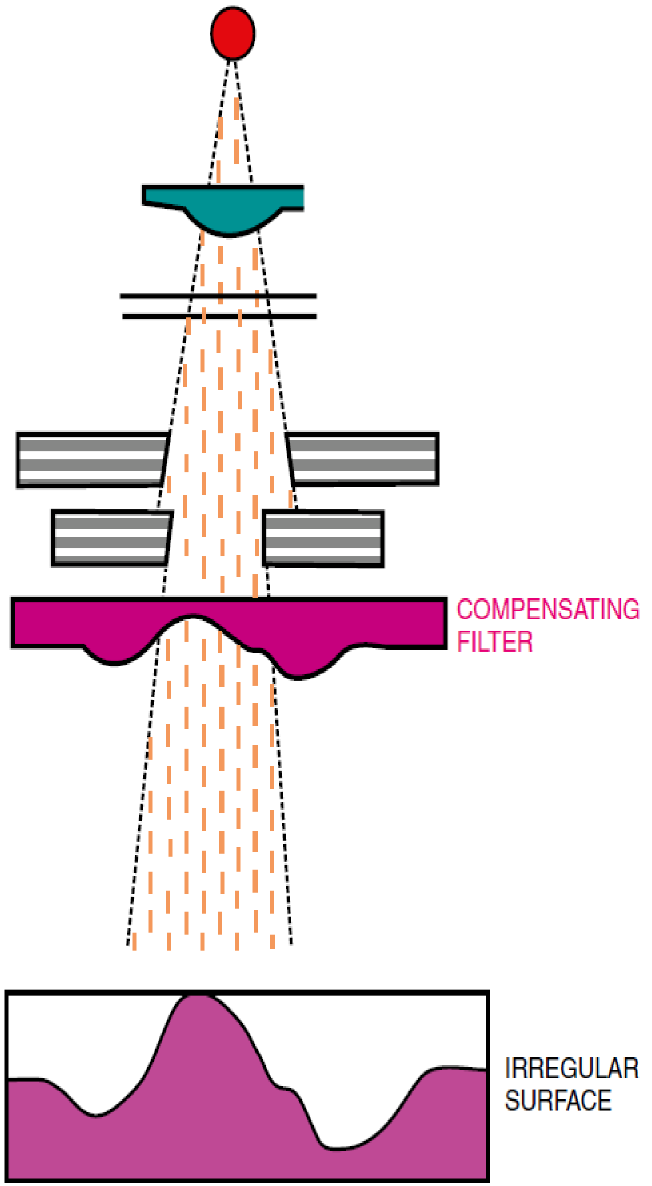
- Bolus is a tissue equivalent material placed directly on the skin surface to even out the irregular patient contour and thereby provide a flat surface for normal beam incidence

- This method suffers a serious drawback: for MV beams, it results in the loss of the skin sparing effect in the skin under the bolus layer



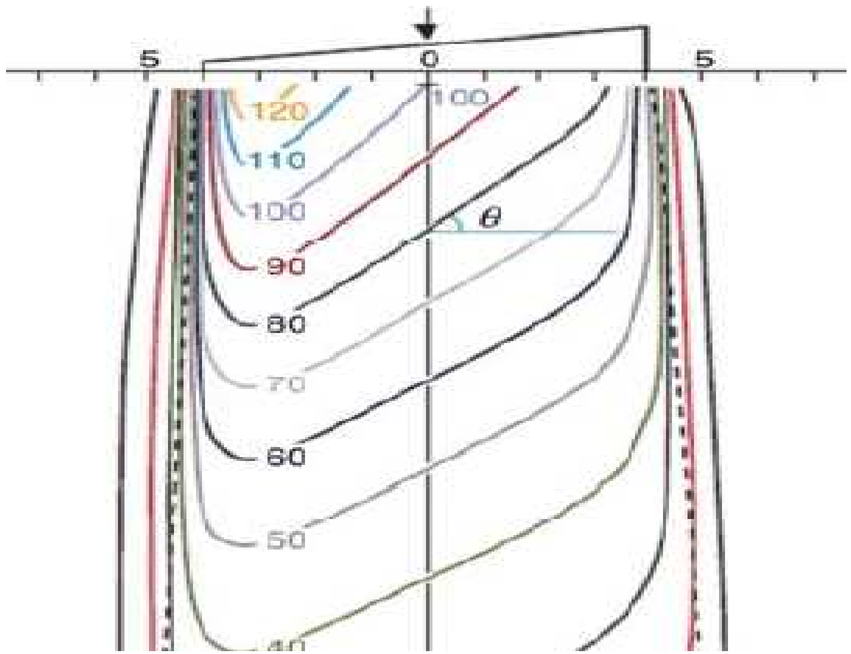
Corrections for irregular contours: Compensators

- Compensators are used to produce the same effect as the bolus yet preserve the skin sparing effect of MV photon beams
- They are custom-made devices that mimic the shape of bolus but are placed in the radiation beam at some distance from the skin surface .
- Usually compensators made of aluminium or brass materials



Corrections for irregular contours: Wedge

- Wedge filters are wedge shaped absorber that causes a progressive decrease in the intensity across the beam, resulting in a tilt of the isodose curves from their normal positions.
- The isodose curves are tilted towards the thin end, and the degree of tilt is depends on the slope of the wedge filter
- Usually wedge is made of dense material such as steel or lead

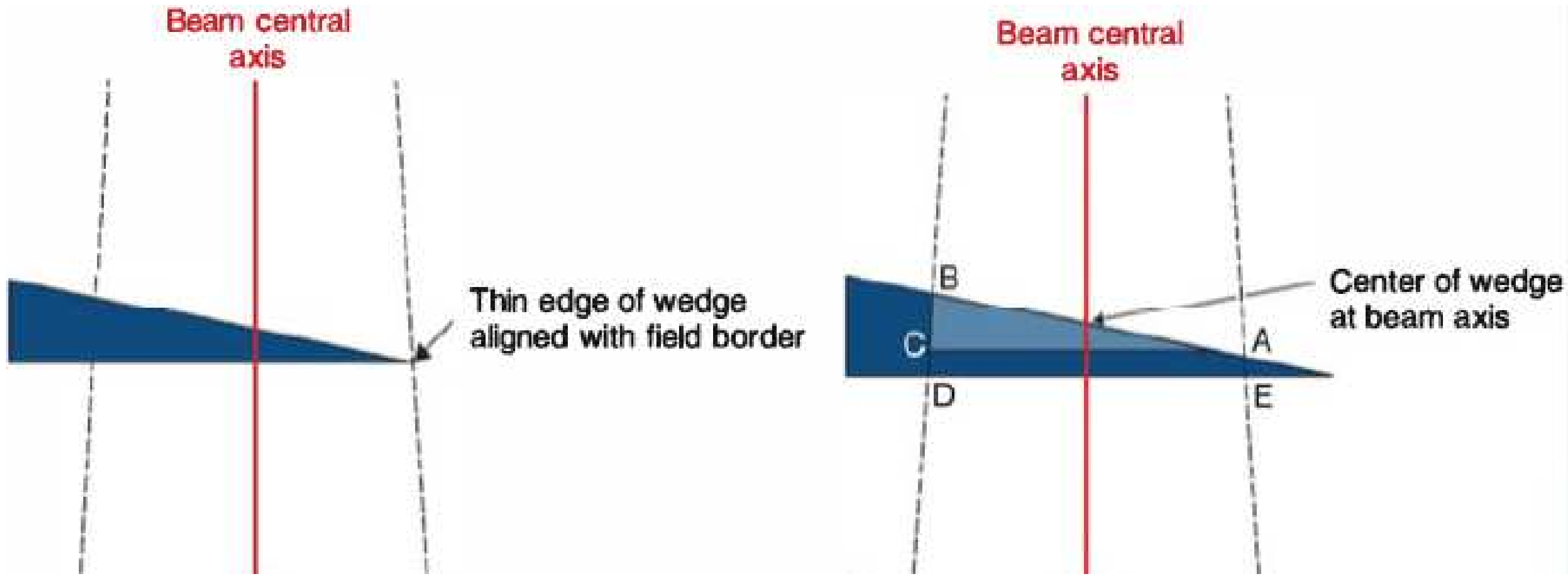


Corrections for irregular contours: Wedge

- Two types of wedges:
    - Physical wedges
    - Dynamic wedges (Dynamic, Motorized, Virtual)
- Dynamic wedges provide the wedge effect on isodose curves through movement of collimator jaws from one end to other end of field
- The advantage of dynamic wedge is automation of treatment delivery and less peripheral dose.
- No beam hardening effect in dynamic wedge
- The main drawback is greater difficulty in radiation dosimetry and beam modeling

Corrections for irregular contours: Wedge

- Physical wedge is usually either
  - Individual wedge – Different wedge for different field size
  - Universal wedge - Single wedge for all field size
- External physical wedges is manually inserted into the beam path
- i.e. Physical wedge frame that can be inserted in the designated slot in the head of the machine
- Usually external wedges are placed at least 50cm from the isocenter to preserve the skin sparing effect of MV beams.



Corrections for irregular contours: Wedge

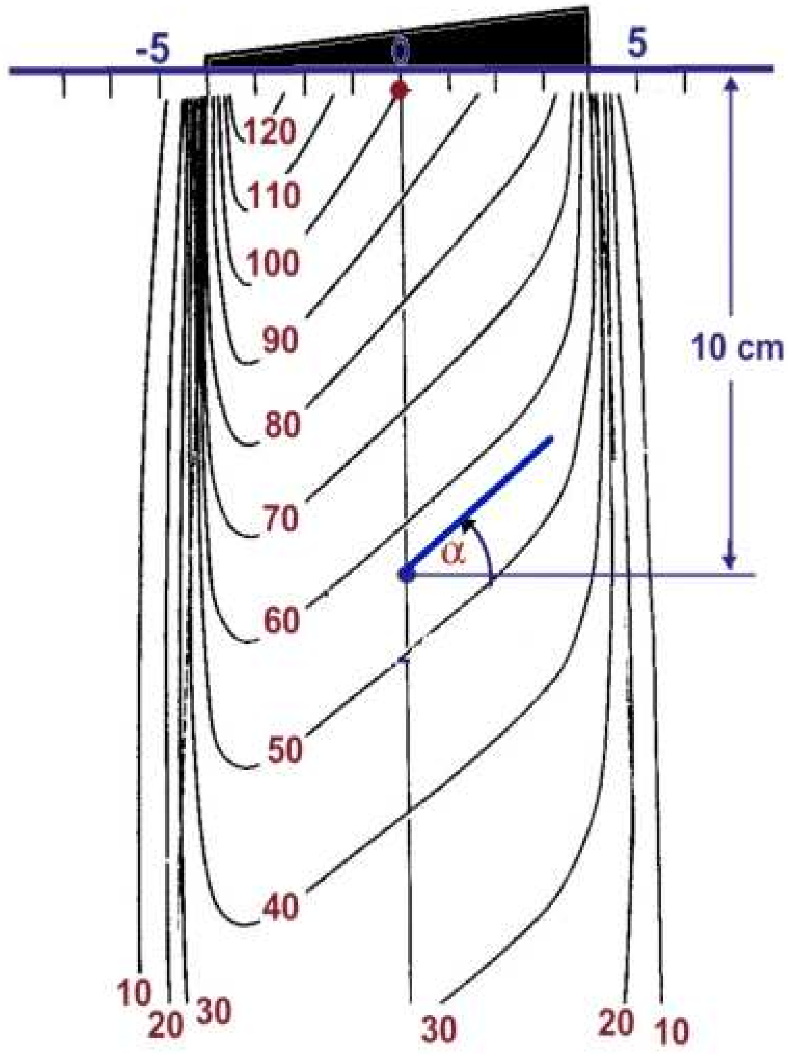
- The presence of a wedge filter decreases the output of the machine.
- This effect is characterized by the wedge transmission factor or wedge factor
- Wedge factor is defined as,

$$WF = \frac{\text{Absorbed dose with wedge}}{\text{Absorbed dose without wedge}}$$

- Usually wedge factor should be measured in phantom at a suitable depth(10cm) beyond the d_{max}

Corrections for irregular contours: Wedge

- Wedge angle refers to " the angle through which an isodose curve is tilted at the central ray of the beam at a specified depth" (tilt by 50% isodose for Co-60 and tilt by isodose curve at 10 cm for MV photons)
- The presence of scattered radiation causes the angle of isodose tilt to decrease with increasing depth in the phantom. Thus a reference depth of 10cm for wedge angle is recommended



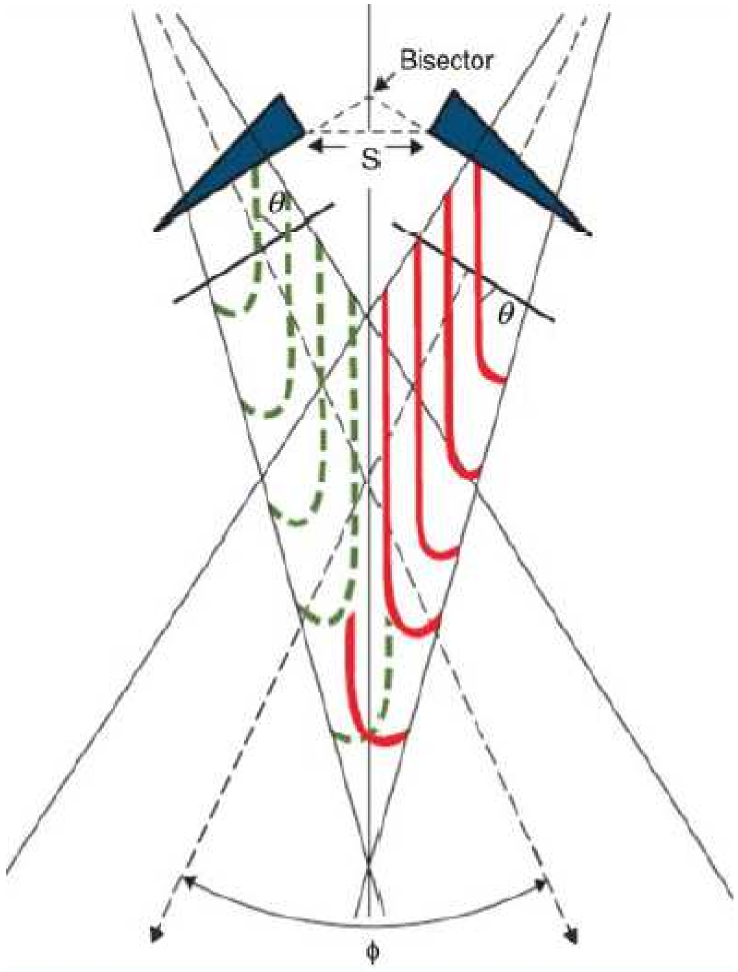
Corrections for irregular contours: Wedge

- Hinge angle : The angle between the central axes of the two beams
- The relation between wedge angle and hinge angle is:

$$\theta = 90 - \Phi/2$$

Here ϕ = hinge angle

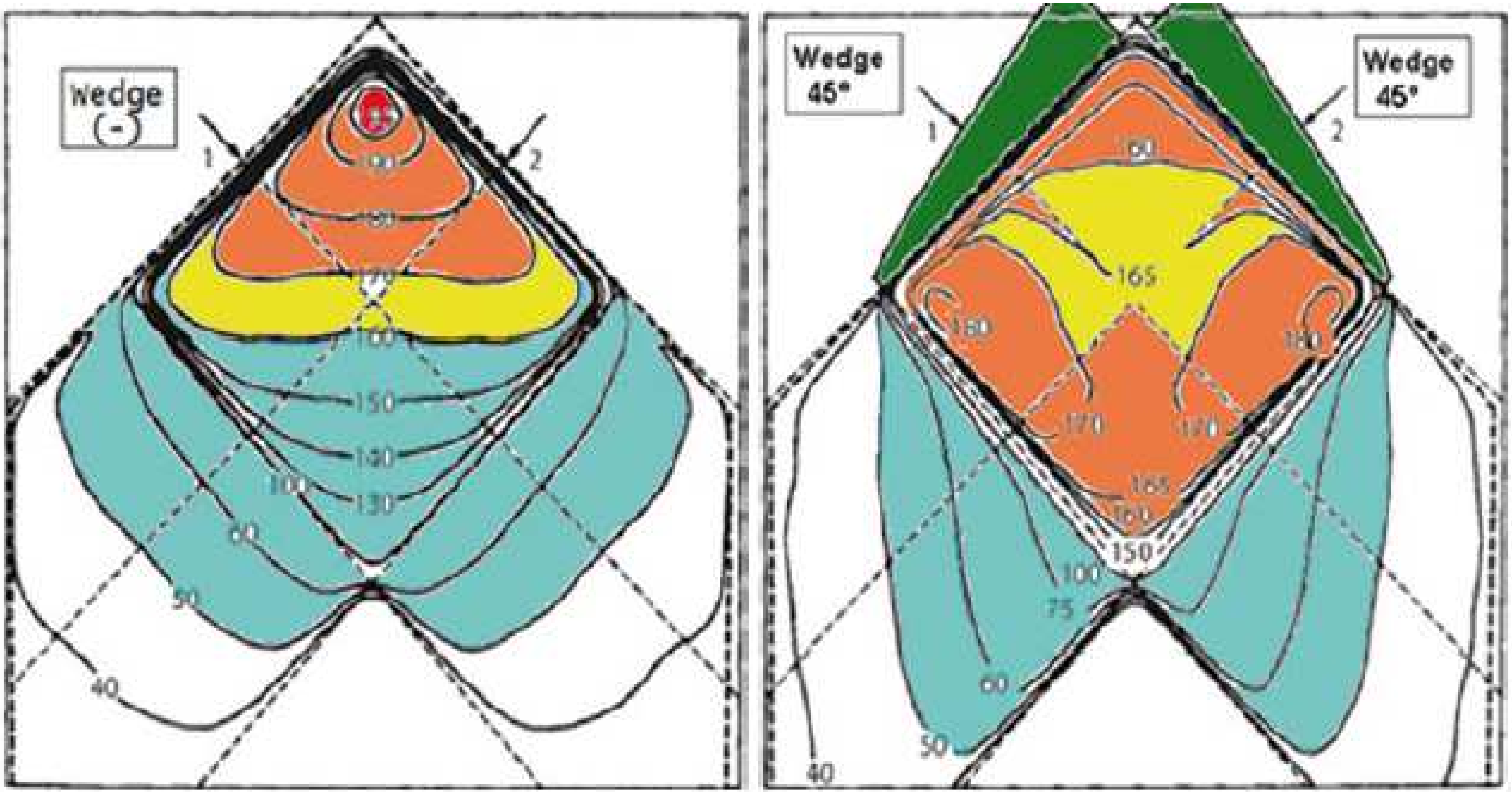
θ = wedge angle



- The determination of wedge angle can be possible by knowing the hinge angle

Corrections for irregular contours: Wedge

The dose distribution before and after use of wedges



Corrections for irregular contours: Blocks

- Shielding of vital organs within a radiation field is one of the major concerns of radiation therapy
- Shielding against primary radiation for superficial and ortho-voltage beams is readily accomplished by thin lead sheets that can be placed or moulded on to the skin surface
- As the beam energy increases to the MV range, thickness of lead required for shielding increases (5 HVL)
- In such conditions, lead blocks are placed above the patient supported in the beam on a transparent plastic tray, called shadow tray.
- The blocks should be shaped or tapered so that their side follow the geometric divergence of the beam. (This minimizes the block transmission penumbra)

Corrections for irregular contours: Blocks

- Blocks may be either standard or custom made
- Standard blocks:
  - Available in various sizes and shapes and designed according to the area that needs to be shielded
  - Mostly standard blocks made of Lead

Corrections for irregular contours: Blocks

Custom blocks:

- Individually made blocks and designed according to the simulation procedure

- Usually made of Cerrobend

- Cerrobend is a mixture of Lead (26.7%), Bismuth (50%), Zinc (13.3%) and Cadmium (10%)

- The main advantage of cerrobend is that it melts at about 70^0C compared with 327^0C for lead and at room temperature it is harder than lead

- Cerrobend density (9.4 g/cc at 20^0C) is lesser than Lead density (11.37 g/cc). Because of this 1.21 times of Cerrobend is equal to 1 time of lead

Corrections for irregular contours: Blocks

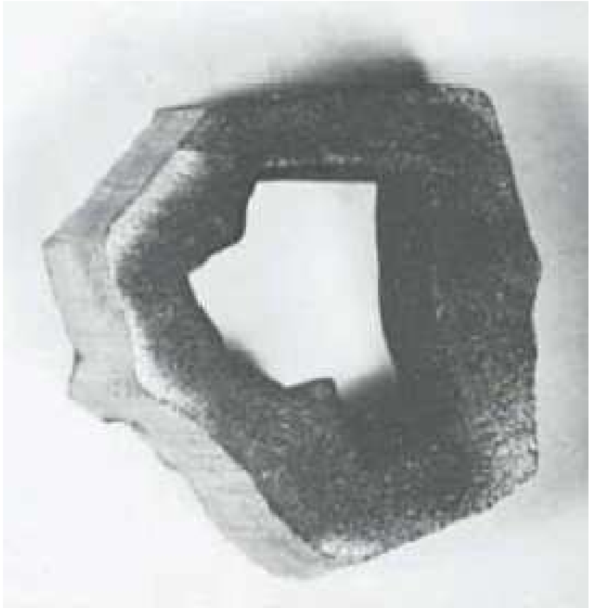
Custom blocks

- Styrofoam cutting device is used to made divergent custom blocks.

- These blocks be either positive or negative

  - Positive : Central portion use to shields
  - Negative: Peripheral area use to blocks

Positive Block



Negative Block

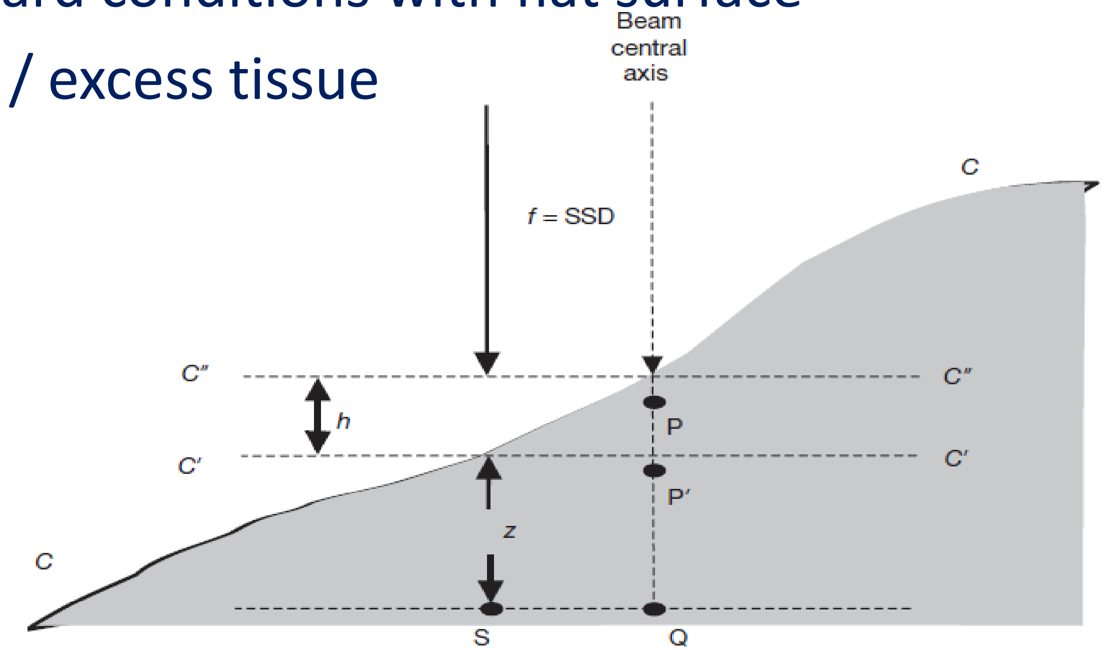
Dose Corrections for irregular contours:

Effective SSD Method

- An assumption is made that the PDD does not depends on the SSD for deviations from the nominal SSD of the order of h (h<<f)
- In this method, the $\mathrm{PDD}_{\mathrm{corr}}$ is determined from:

$$PDD_{corr} = PDD' \times \left(\frac{f + d_{\max}}{f + h + d_{\max}}\right)^2$$

  - $\mathrm{PDD}^{/}$ = PDD under standard conditions with flat surface
  - h = Thickness of missing / excess tissue
  - f = SSD

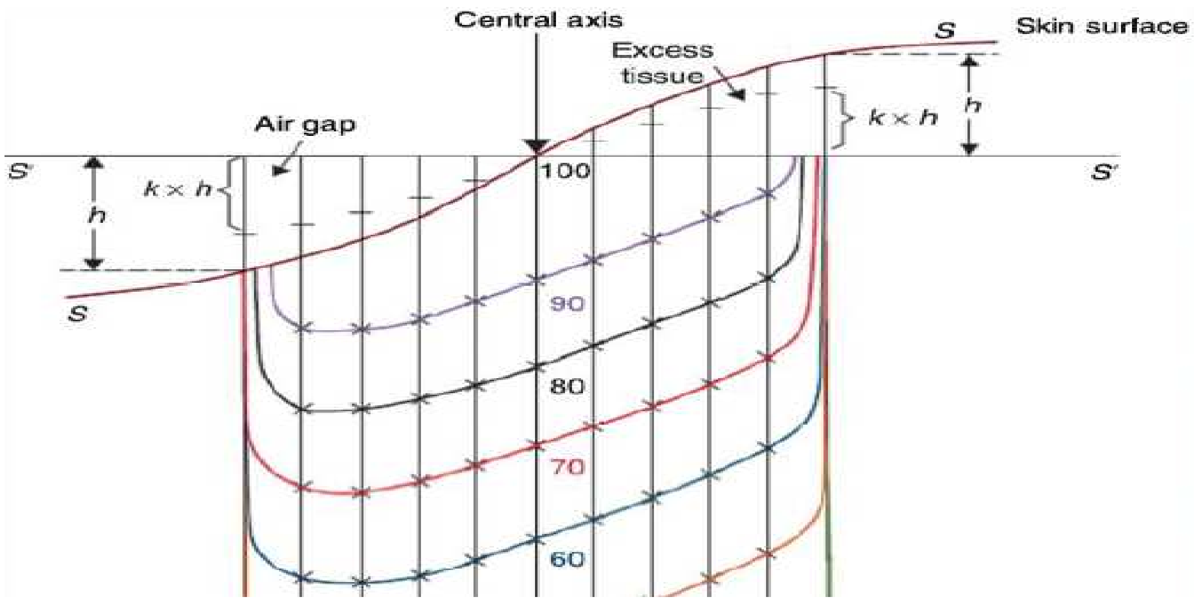


Dose Corrections for irregular contours:

Isodose Shift Method

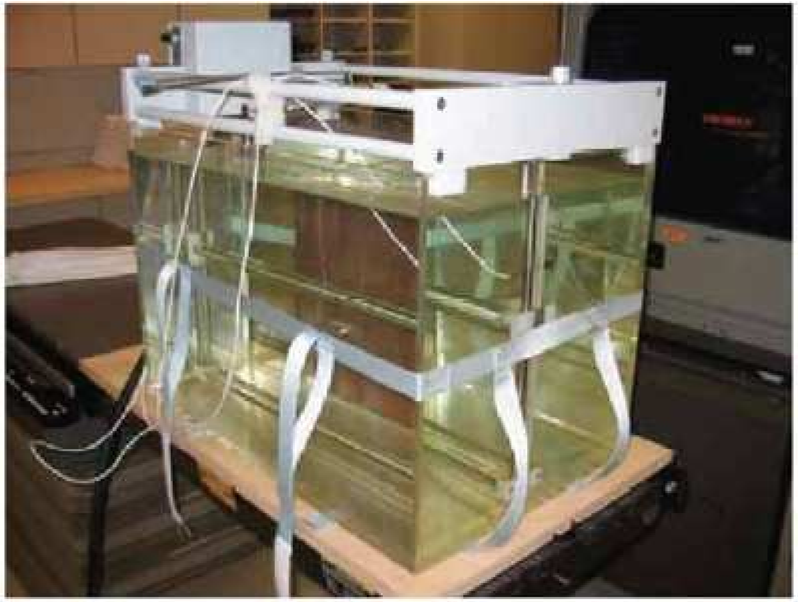
- In early method, useful for individual point dose calculations.
- In this method, the entire isodose line is shifted either upwards or downwards by (h x k), where h is the thickness of the missing or excess tissue and k is a factor depending on beam energy
- For missing tissue, h is positive and the isodose is shifted away from the source,
- h is negative for excess tissue and the isodose is shifted towards the source

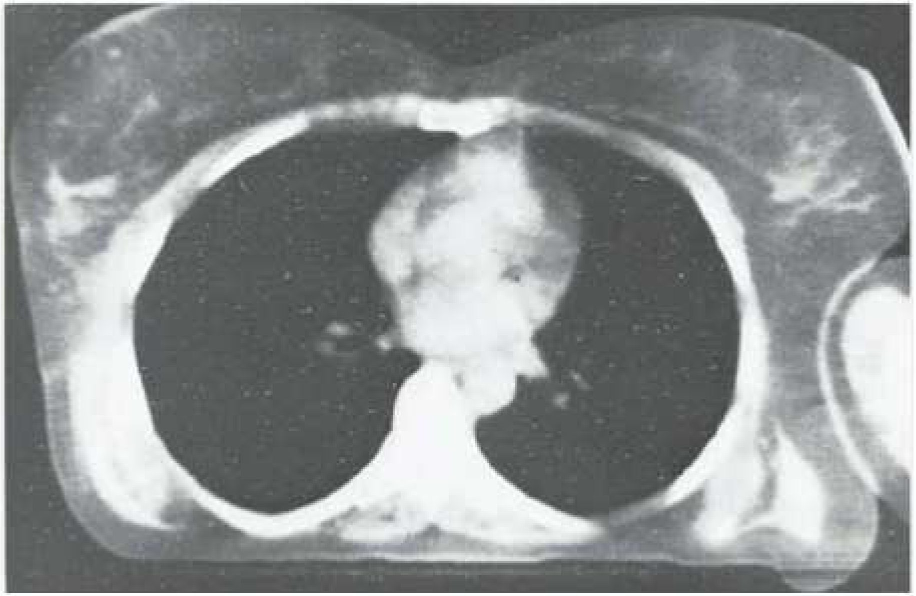
Photon Energy	K value
Up to 1MV	0.8
60Co – 5MV	0.7
5- 15MV	0.6
15 – 30	0.5



Corrections for Tissue Inhomogeneities

- Patient body consists of various tissues that may differ from water in density and atomic number
- Standard isodose charts and PDD data are given for uniform density water phantoms
- The direct application of standard isodose curves & PDD results inhomogeneous dose distribution





Dose Corrections for Tissue Inhomogeneities

- The tissue inhomogeneity
    - Increase or decrease in the attenuation of the primary beam, which affects the distribution of the scattered radiation
    - Increase or decrease of the secondary electron Fluence
- Empirical methods are available for correcting the water phantom dose to tissue doses
    - TAR method
    - Power Law TAR method
    - Isodose shift method

- TAR Ratio method: Uses the radiographic depth to calculate a new TAR

- Batho Power law method: calculates TAR based on an exponential function of depth

Dose Corrections for Tissue Inhomogeneities: Isodose shift method

- This method is used for manually correcting isodose curves for the presence of inhomogeneities
- The isodose curves beyond the inhomogeneity are moved by an amount equal to n times the thickness of the inhomogeneity
- The shift is towards the skin for bone and away from the skin for lung or air cavity
- The n values for ^{60}Co and 4MV x-ray are experimentally determined. These factors are independent of field size.

Inhomogeneity	Shift factor (n)
Air cavity	-0.6
Lung	-0.4
Hard bone	0.5
Spongy bone	0.25

SSD Setup

- SSD setup uses a constant distance between source and the surface

- SSD can be changed as needed (80, 100, 110 cm etc.,)
- Increasing the depth of the prescription point will increase its distance from the source

- PDD is used for SSD dose calculations

SAD Setup

- SAD setup uses a constant distance between the source and isocenter
- In SAD technique, the radiation source moves in a circle around the axis of rotation, which is usually placed in the tumor.
- Although SSD varies depending on the shape of the surface contour, the SAD remains constant
- This allows for rotation around a fixed isocenter, and is therefore much more common for modern-era radiation therapy
- SAD is a fixed value for any given machine (80cm for Co-60, 100cm for Linac)
- TAR/TPR/TMR are used for SAD dose calculations

Treatment Time Calculation

$$\text{Treatment Time} = \frac{\text{Dose}}{\text{O/P} * \text{Sc} * \text{Sp} * \text{PDD/TMR} * \text{WF} * \text{TF} * (\text{ISL Factor})}$$

O/P- Treatment Unit output (CGy/min or CGy/MU)

S_c-Collimator Scatter Factor

S_p-Phantom Scatter Factor

PDD-Percentage Depth Dose

TMR-Tissue Maximum Ratio

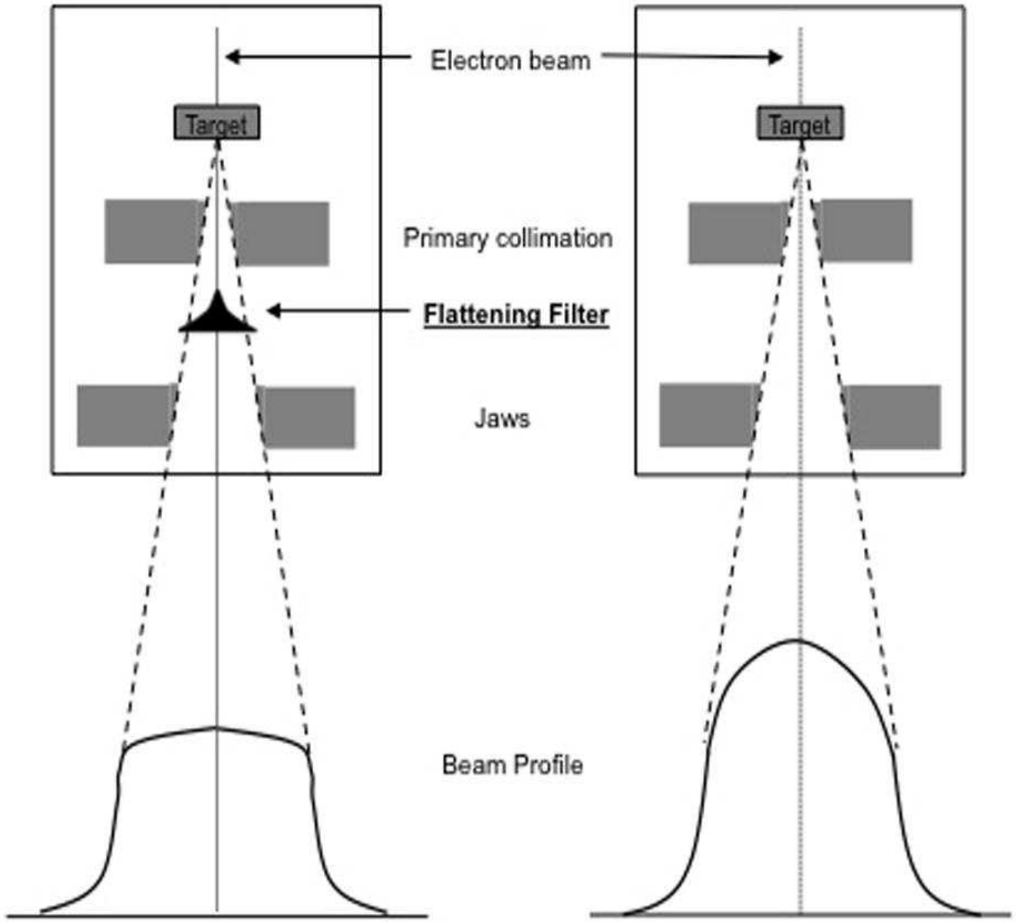
WF – Wedge Factor

TF-Tray factor

ISL Factor – Inverse Square Law factor

Flattening Filter Free(FFF) Beam

Flattening Filter converts the forward peaked MV x-ray photon intensity into uniform intensity pattern



Linac with flattening filter, corresponding dose profile

Linac without flattening filter, corresponding dose profile

Advantage of FFF beam

- Significant increase in dose rate by a factor of about 2-4 times
- Softening of the x-ray spectra
- Leading to reduction in scattered radiation
- Reduction in neutron and photon leakage from the treatment head

FFF beam parameters

Following dosimetric parameters for FFF beam differs from FF beam

- Field size definition
- Beam quality
- Surface dose
- Off axis ratio (OAR)
- Beam flatness
- Symmetry
- Penumbra
- Depth dose profiles

The characteristics of flat photon beams are not applicable to unflat photon beams because of significant differences in shape of their beam profiles

FFF beam of various accelerators

TABLE 1. Characteristics of commercially available FFF beams. All dosimetric quantities are given for a 10×10 cm^2 field at 100 cm SSD unless otherwise noted and were provided by the manufacturers.

	Varian		*Elekta*		*Siemens*			
Nominal energy (MV)	6 FFF	10 FFF	6 FFF	10 FFF	7 UF	11 UF	14 UF	17 UF
Bremsstrahlung target material	Tungsten		Tungsten		Tungsten			
Approximate mean electron energy on target (MeV)	6.2	10.5	7	10.5	8.9	14.4	16.4	18.3
Filtration	0.8 mm Brass		2mm Stainless steel		1.27 mm Al			
d_{max} (cm)	1.5	2.3	1.7	2.4	1.9	2.7	3.0	3.3
Dose at 10 cm depth (%)	64.2	71.7	67.5	73.0	68.5	74.5	76.5	78.0
Dose 10 cm from central axis (40×40 cm^2 field), at d_{max} (%)	77	60	70[a]	59[a]	68	57	-	-
Maximum dose rate on beam axis at d_{max} (cGy/min)	1400	2400	1400	2200	2000	2000	2000	2000
Dose per pulse on beam axis at d_{max} (cGy/pulse)	0.08	0.13	0.06	0.09/0.14[b]	0.13	0.13	0.13	0.13

TG- 100 Recommendations

AAPM TG -100 recommends the generation of following data sets/parameters

- Beam energy
- Surface dose
- Field size, flatness, symmetry, and penumbra

Analysis of FFF beam

OFF AXIS RATIO

- The off-axis ratio at ± 3 cm lateral distance from central axis at 10 cm depth for 10 cm × 10 cm collimator setting

DEPTH DOSE PROFILES

- Depth dose profiles for 5 cm × 5 cm, 10 cm × 10 cm and 20 cm × 20 cm collimator setting
- The depth of maximum dose and percentage depth dose at 10 cm depth

BEAM PROFILES

- Beam profiles for 20 cm × 20 cm collimator setting at 10 cm depth in isocentric setup for all the available unflattened photon beam energies shall be measured

Analysis of FFF beam

Field size

- The constancy of the beam profiles along major axes is verified by the mid separation between inflection points (IPs).

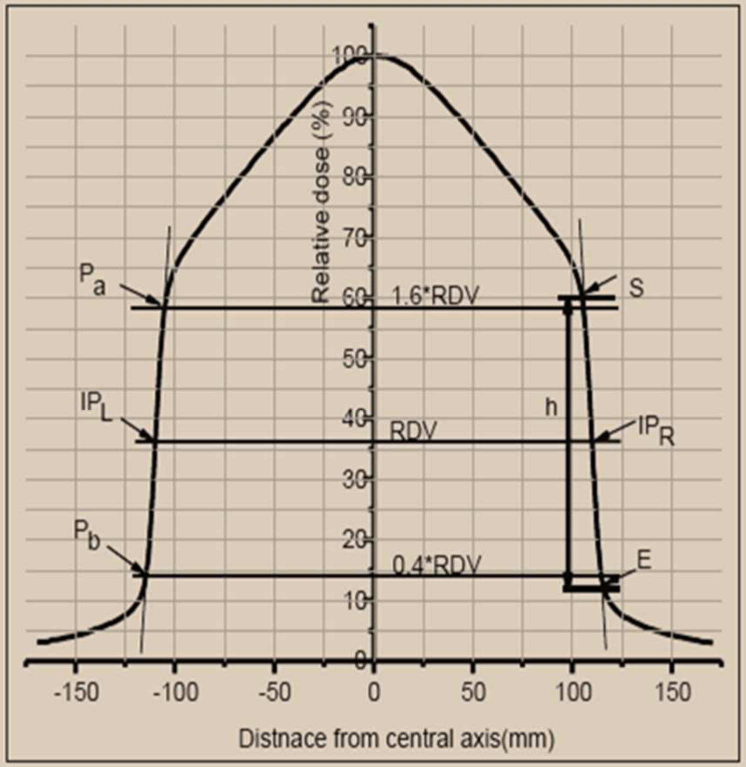


Figure 1: Schematic diagram for determining inflection point and penumbra

Analysis of FFF beam

Symmetry

- Symmetry is evaluated by International Electrotechnical Commission (IEC 60976, 2008) report

Degree of unflatness

- It is quantified by the lateral distance from the central axis at 90%, 75% and 60% dose points on either side of the beam profile

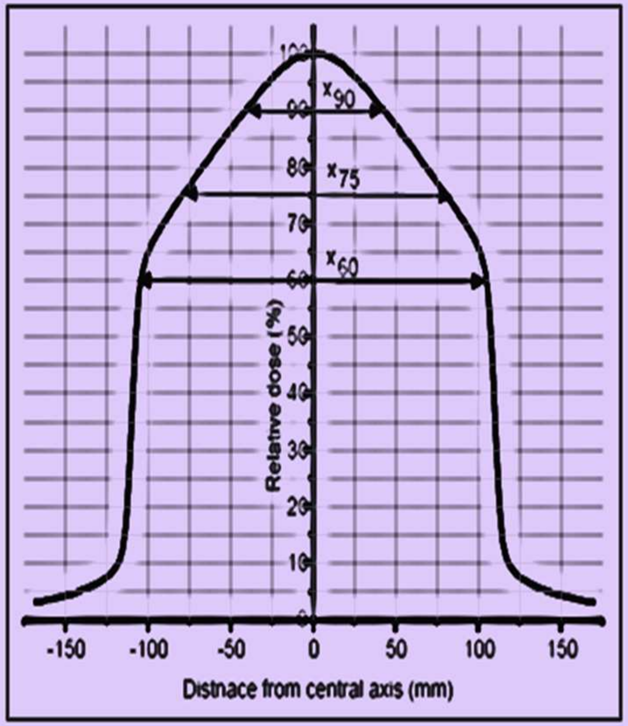


Figure 2: Schematic diagram for determining degree of unflatness

Analysis of FFF beam

Penumbra

- To determine penumbra, dose value at mid IP shall be taken as reference dose value (RDV)
- Points Pa and Pb, are located at 1.6 and 0.4 times of RDV, respectively
- Lateral separation between Pa and Pb on either side of the profile will be the measure of the radiation beam penumbra.

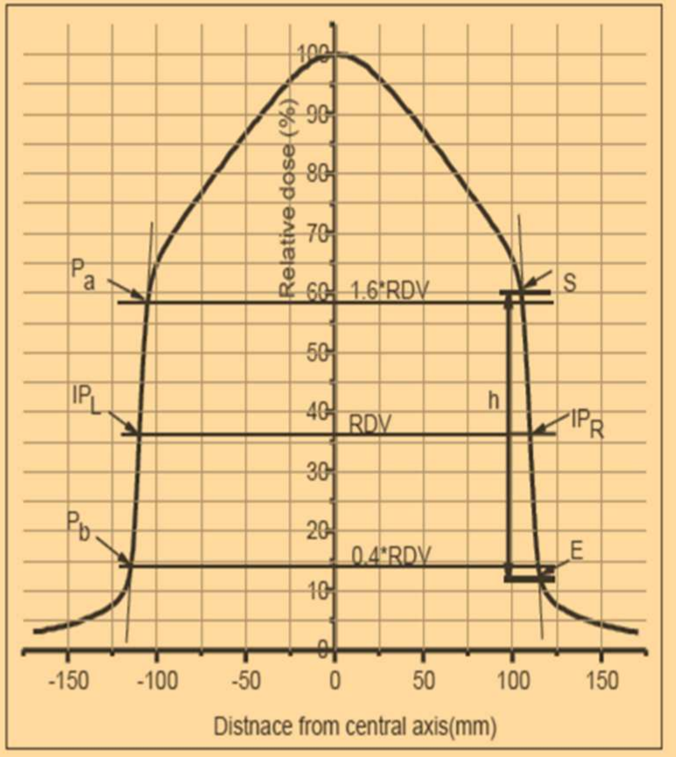


Figure 1: Schematic diagram for determining inflection point and penumbra

Periodic QA Tests

- Periodic QA tests should be carried out on daily/monthly basis and proper records should be maintained

- The constancy in the performance of the linac in comparison to baseline data of the given parameter generated at the time of acceptance testing/clinical commissioning should be verified

Periodic QA Tests

Energy check

- The TPR20/10 should be measured for 10 cm x 10 cm collimator setting

Measurements of OAR

- The OAR should be measured at ± 3 cm for 10 cm x 10 cm collimator setting

Measurements of profiles

- The beam profiles should be measured using multiple detector system/any other suitable device for 20 cm x 20 cm collimator setting

Thank you
for your
Kind attention

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