Basic Radiation Oncology Physics

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Acknowledgment:

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Entire credit belongs to IAEA.

Penetration of photon beams into patient

- A photon beam propagating through air or vacuum is governed by the inverse square law.
- A photon beam propagating through a phantom or patient is affected not only by the inverse square law but also by the attenuation and scattering of the photon beam inside the phantom or patient.
- The three effects make the dose deposition in a phantom or patient a complicated process and its determination a complex task.

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Need for dosimetric functions

- We need to know dose distribution within target volume and in the surrounding tissues
- One cannot measure dose at all points
- Several dosimetric functions link the dose at any arbitrary point inside the patient to the known dose at the beam calibration (or reference) point in a phantom.

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Dosimetric functions are measured in reference conditions

- Dosimetric functions are usually measured with suitable radiation detectors in tissue equivalent phantoms.
- Dose or dose rate at the reference point is determined for, or in, water phantoms for a specific set of reference conditions, such as:
  - Depth in phantom *z*
  - Field size *A*
  - Source-surface distance (SSD)

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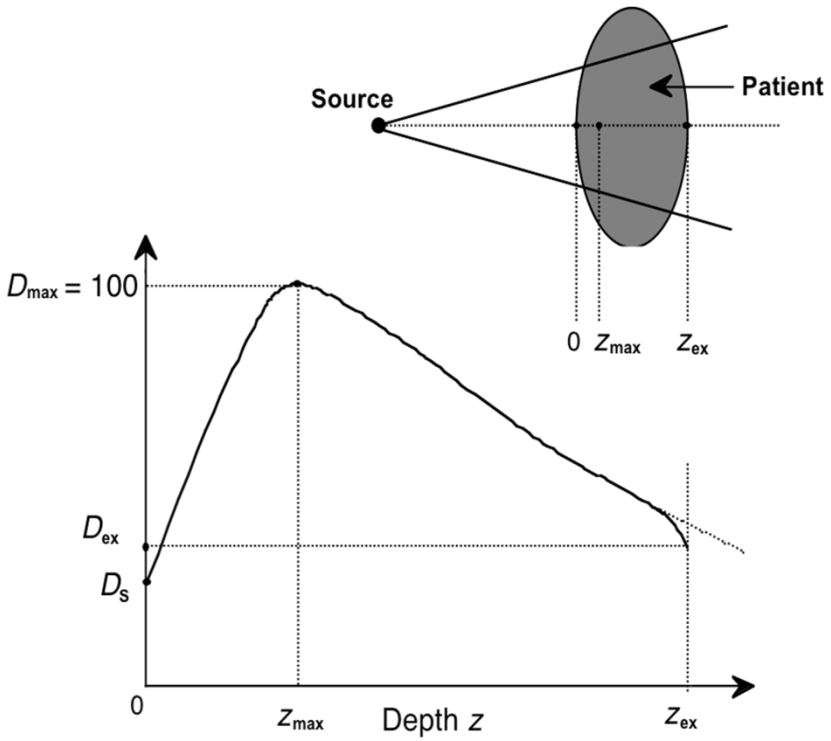
Central axis depth dose (CADD)

- Typical dose distribution for an external photon beam follows a known general pattern:
  - Surface dose D_s
  - Rapid buildup of dose beneath the surface - reaches a maximum value at a depth z_{max} - decreases almost exponentially - reaches a value D_{ex} at the patient's exit point.



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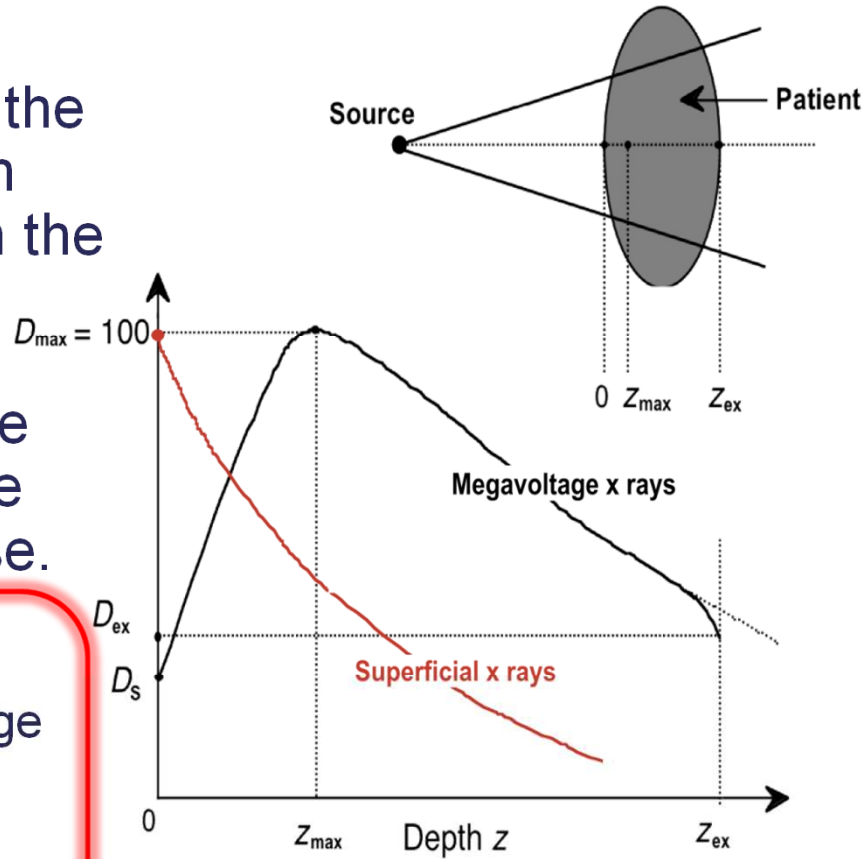
Surface dose

- Surface dose:
  - For megavoltage x-ray beams the surface dose is generally much lower (skin sparing effect) than the maximum dose at z_{max}.
  - For superficial and orthovoltage beams z_{max} = 0 and the surface dose equals the maximum dose.
  - Typical values of surface dose:
    - 100%for superficial and orthovoltage
    - 30% for cobalt-60 gamma rays
    - 15% for 6 MV x-ray beams
    - 10% for18 MV x-ray beams



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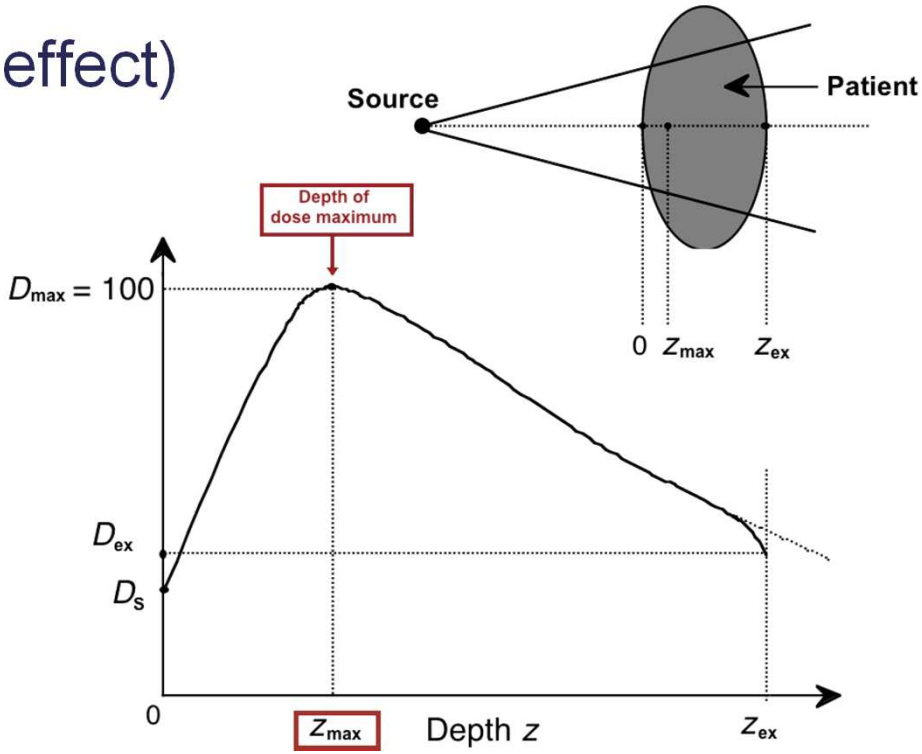
Depth of dose maximum depends on….

- Photon beam energy (main effect)
- Field size (secondary effect)
- For a given field size:
  - z_{max} increases with photon beam energy.
  - For 5x5 cm^2 fields, the nominal values of z_{max} are:



Energy	100 kV_p	350 kV_p	Co-60	4 MV	6 MV	10 MV	18 MV
z_{max}*(cm)*	0	0	0.5	1.0	1.5	2.5	3.5

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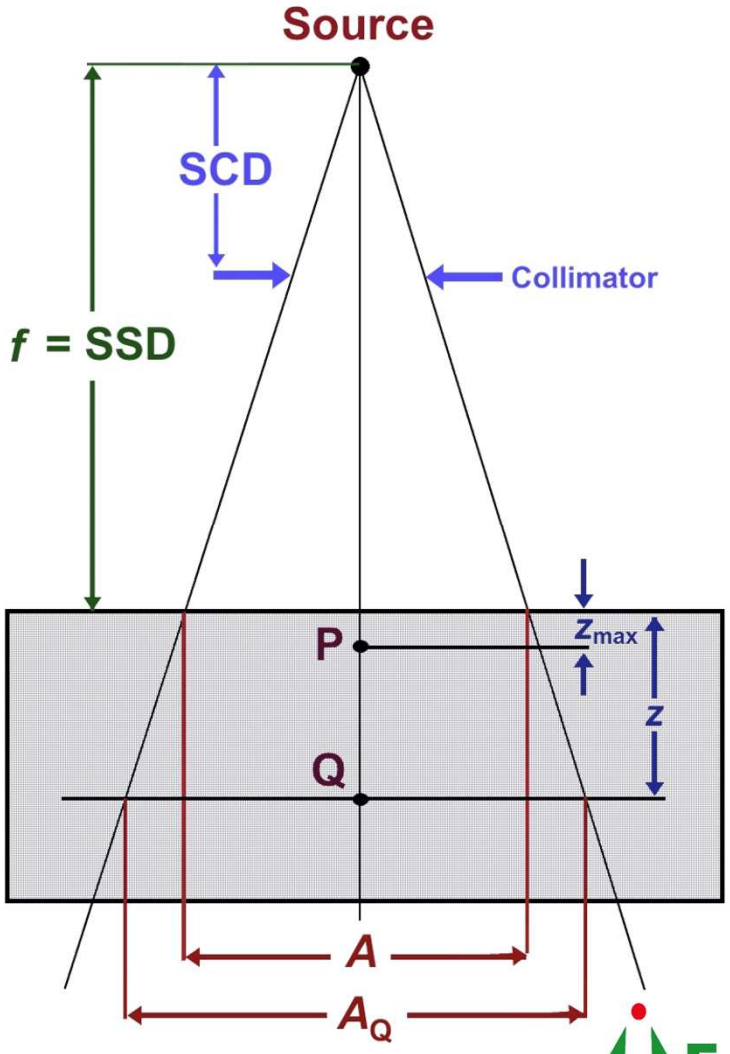
Radiation treatment parameters

- The **main parameters** in external beam dose delivery with photon beams are:
  - Depth of treatment *z*
  - Fields size *A*
  - Source-skin distance (SSD) in SSD setups
  - Source-axis distance (SAD) in SAD setups
  - Photon beam energy
  - Number of beams used in dose delivery to the patient
  - Treatment time for orthovoltage and teletherapy machines
  - Number of monitor units (MUs) for linacs

Radiation treatment parameters

- Point P is at z_{max} on central axis.
- Point Q is arbitrary point at depth z on the central axis.
- Field size A is defined on patient's surface.
- A_Q is the field size at point Q.
- SSD = source-skin distance.
- SCD = source-collimator distance



Radiation field size

- Radiation fields are divided into two categories:

  geometric and dosimetric (physical).

  - According to the ICRU, the geometric field size is defined as "the projection of the distal end of the machine collimator onto a plane perpendicular to the central axis of the radiation beam as seen from the front center of the source."

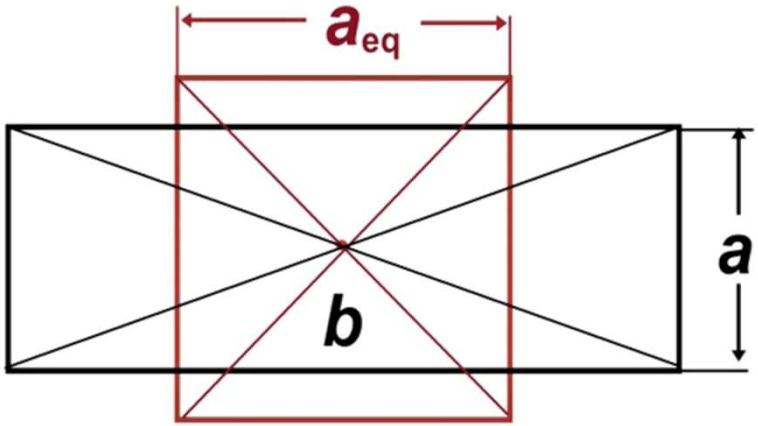
  - The dosimetric field size (also called the physical field size) is defined by the **intercept of a given isodose surface (usually 50%)** with a plane perpendicular to the central axis of the radiation beam at a defined distance from the source.

Equivalent field size

- 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).

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

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



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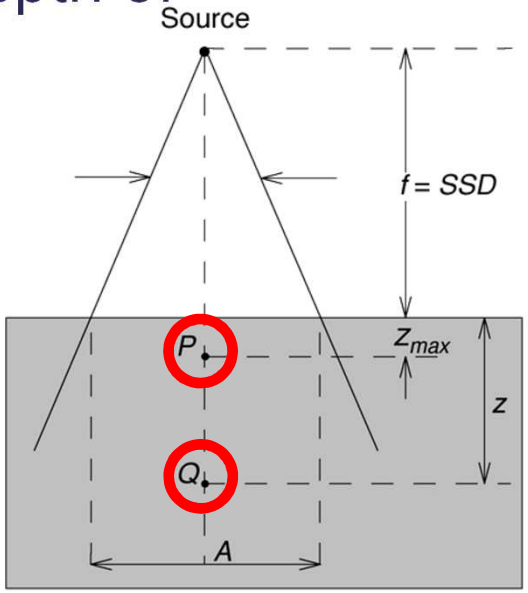
Percentage depth dose

- Central axis dose distributions inside the patient are usually normalized to D_{max} = 100% at the depth of dose maximum z_{max} and then referred to as percentage depth dose (PDD) distributions



- PDD is thus defined as follows:

$$\mathrm{PDD}(z,A,f,h\nu) = 100\frac{D_Q}{D_P} = \frac{\dot{D}_Q}{\dot{D}_P}$$

  - D_Q and $\dot{D}_Q$ are the dose and dose rate, respectively, at arbitrary point Q at depth z on the beam central axis.
  - D_P and $\dot{D}_P$ are the dose and dose rate, respectively, at reference point P at depth z_{max} on the beam central axis.

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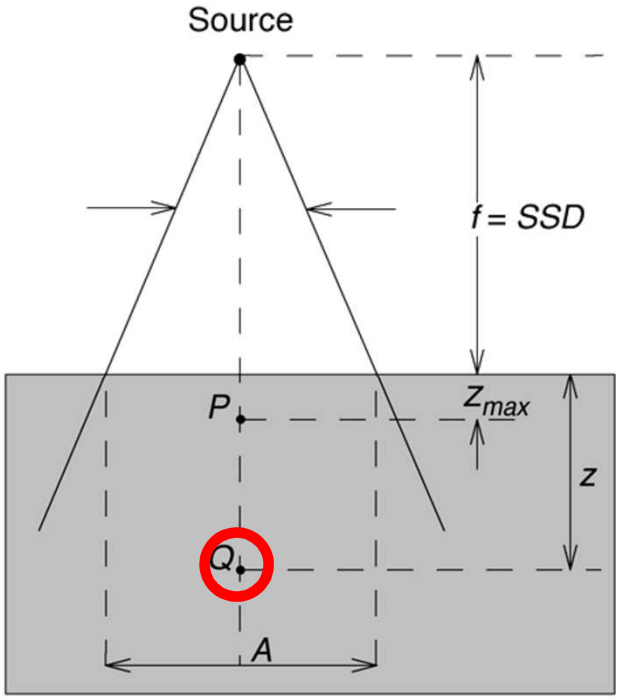
Dose at any point is due to both primary & scatter

- The dose at point Q in the patient consists of two components: **primary component** and **scatter component.**

- $D_Q = D_{pri} + D_{sca}$
  - As the depth increases, the **relative** contribution of D_{pri} decreases and that of D_{sca} increases
  - At low energies, this effect is predominant



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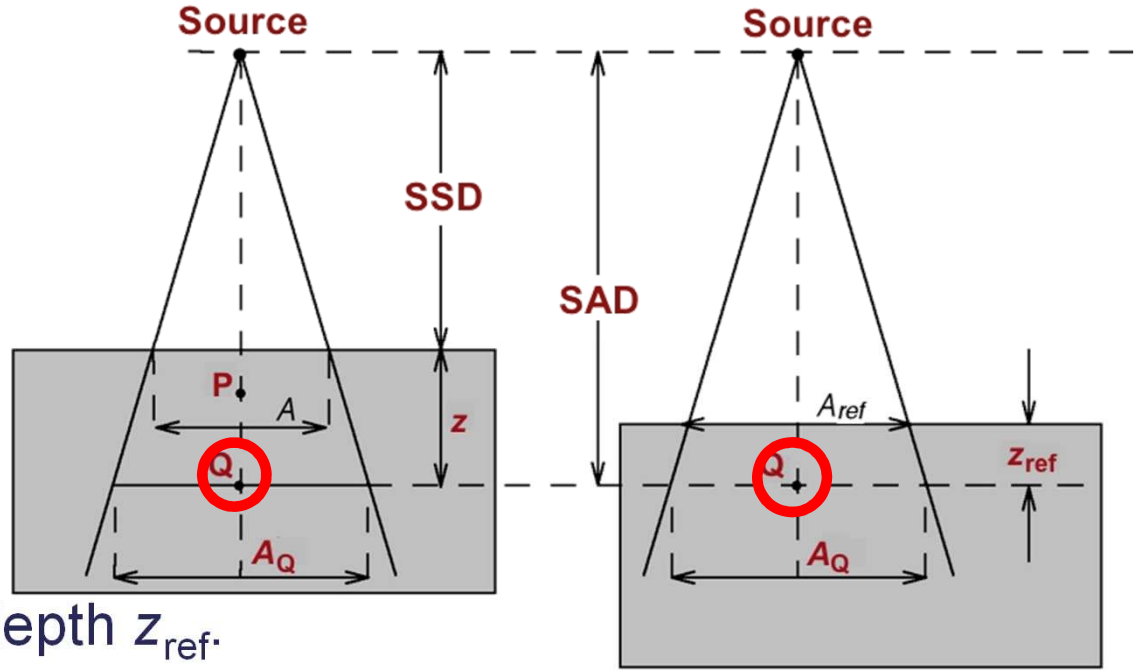
Tissue-phantom ratio (TPR)

- For isocentric setups with megavoltage photon energies the concept of tissue-phantom ratio TPR was developed.
- Similarly to TAR the TPR depends upon z, A_Q, and energy.
- TPR is defined as:

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

  - D_Q is the dose at point Q at depth z
  - D_{Qref} is the dose at depth z_{ref}.



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

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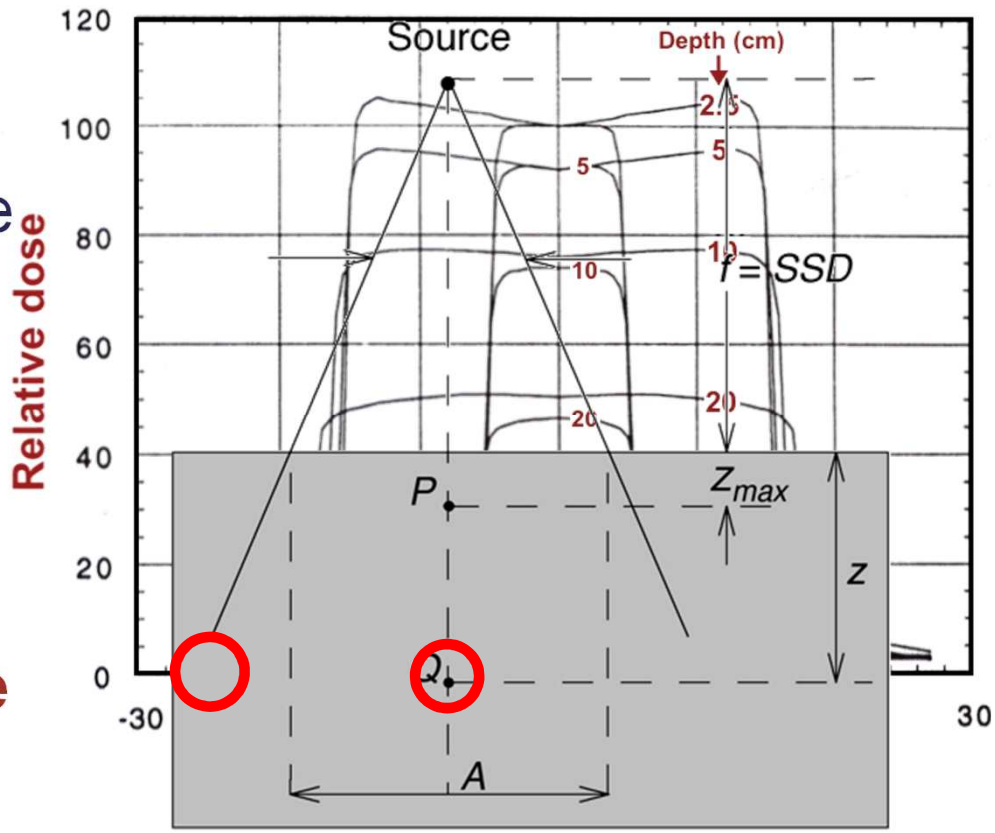
Off-axis beam profiles

- Combining a central axis dose distribution with off-axis data results in a volume dose matrix.
- 2-D and 3-D information on the dose distribution in the patient
- The off-axis ratio OAR is 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



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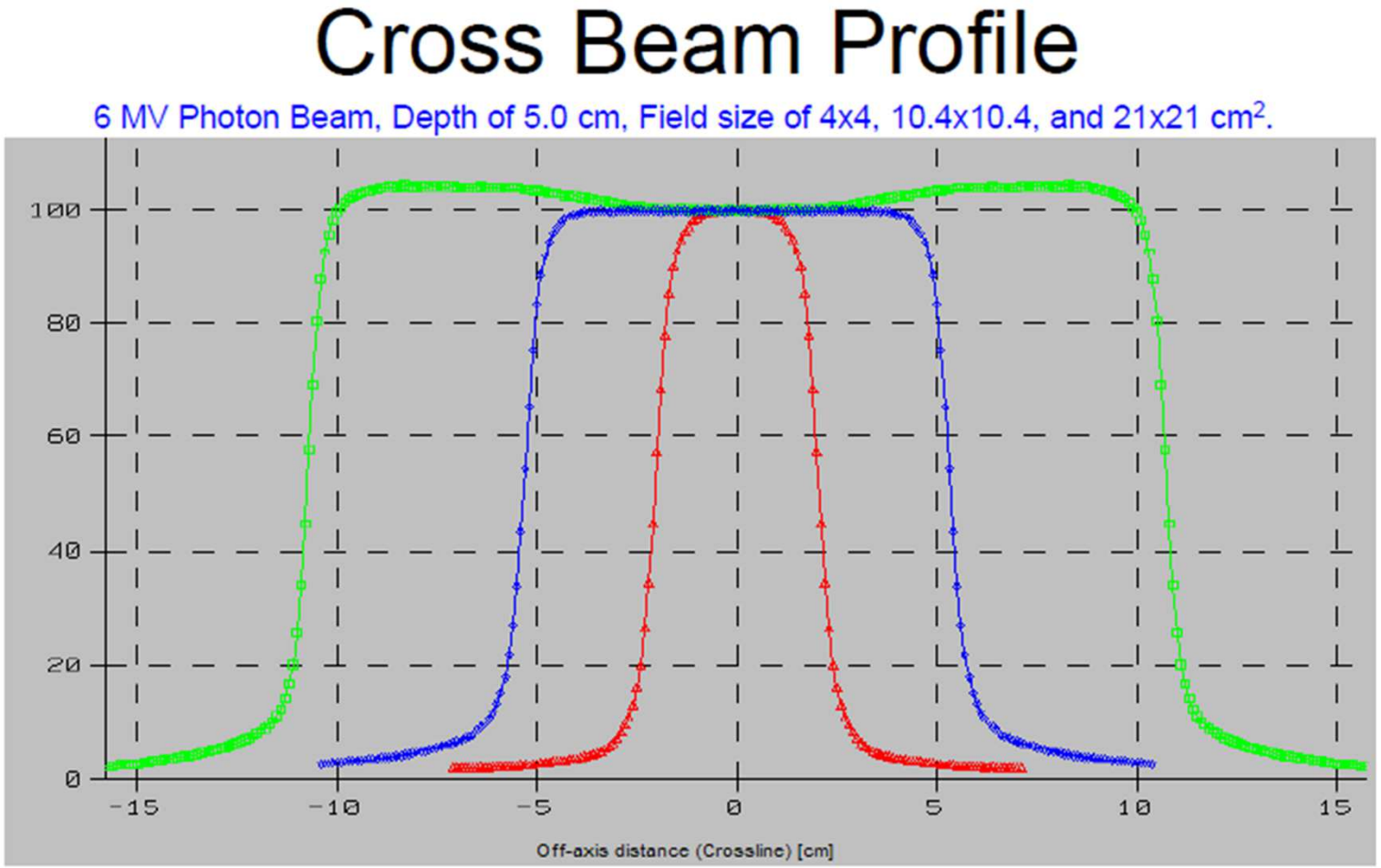
Off-axis or cross-beam beam profiles



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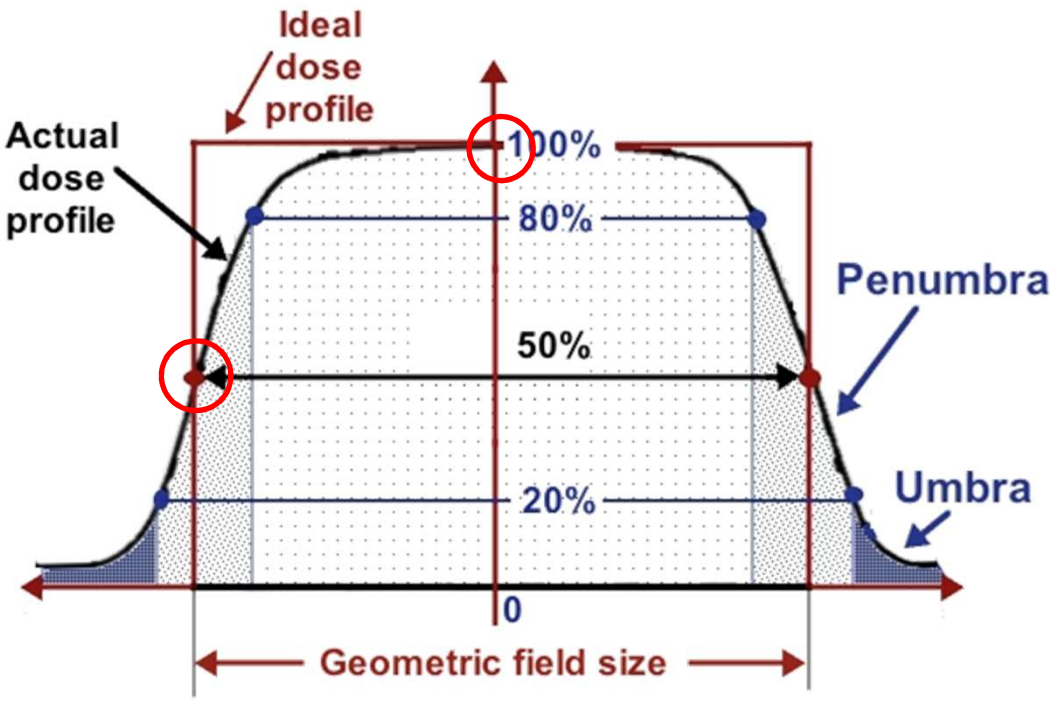
Field size

- Geometric or nominal field size is:
  - Indicated by the optical light field of the treatment machine.
  - Usually defined as the separation between the 50% dose level points on the beam profile measured at the depth of dose maximum z_{max} (dosimetric field size)



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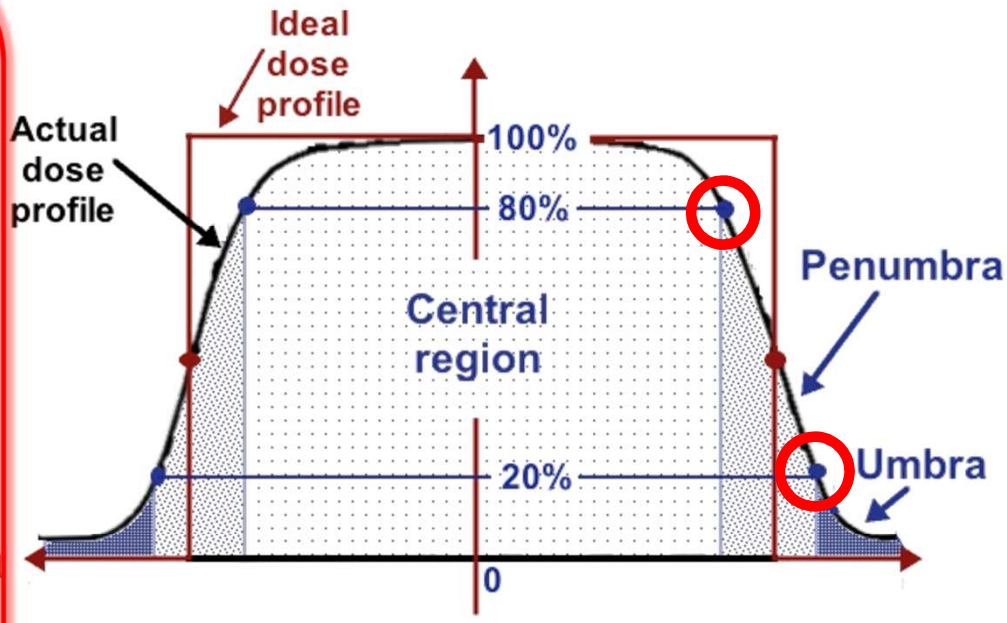
Penumbra

- The total penumbra is referred to as the **physical penumbra** and consists of three components:
  - Geometric penumbra results from the finite source size.
  - Scatter penumbra results from in-patient photon scatter originating in the open field.
  - Transmission penumbra results from beam transmitted through the collimation device.



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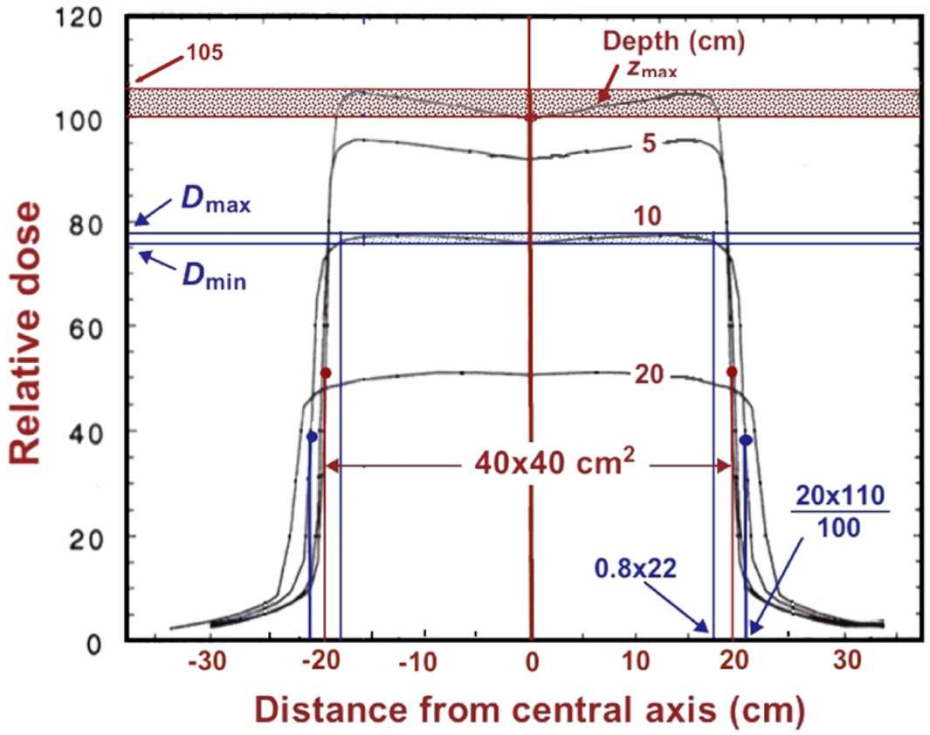
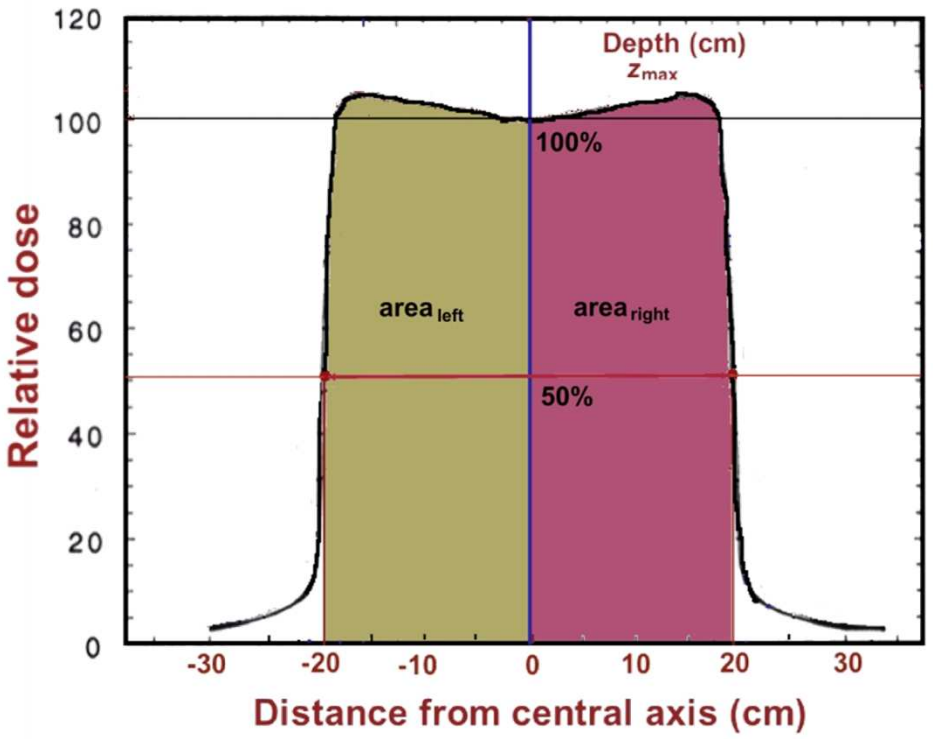
Flatness & Symmetry





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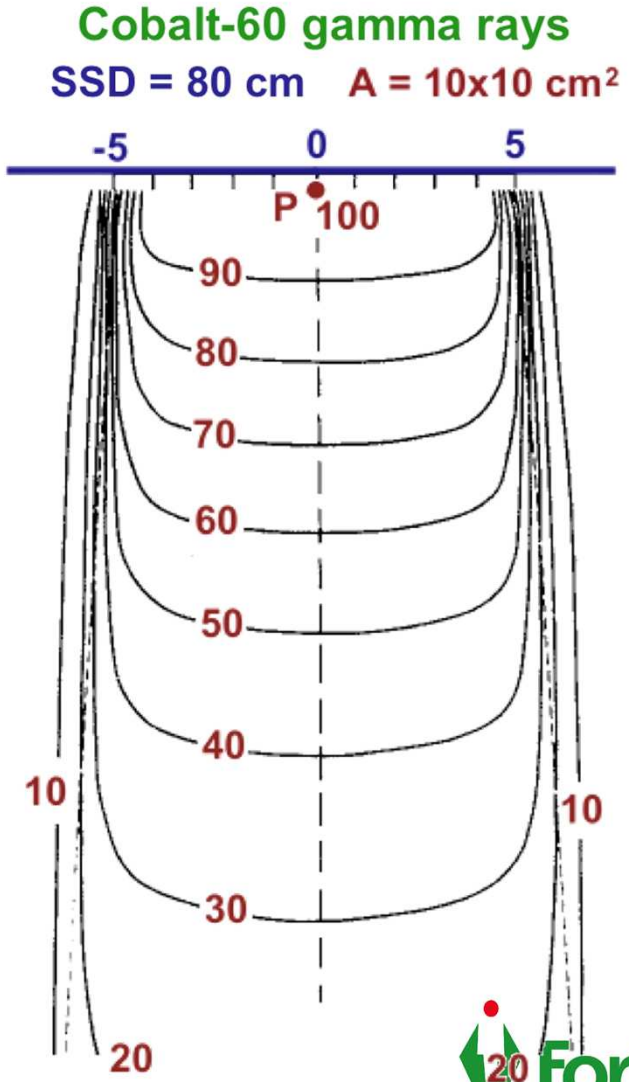
Isodose chart & isodose curves

- An isodose chart for a given single beam consists of a family of isodose curves usually drawn at regular increments of PDD.
- Two normalization conventions are in use:
  - For SSD set-ups, all isodose values are normalized to 100% at point of dose maximum on the central beam axis.
  - For SAD set-ups, the isodose values are normalized to 100% at the isocentre

Normalization at dose maximum

- For SSD set-ups, all isodose values are normalized to 100% at point P on the central beam axis (point of dose maximum at depth z_{max}).



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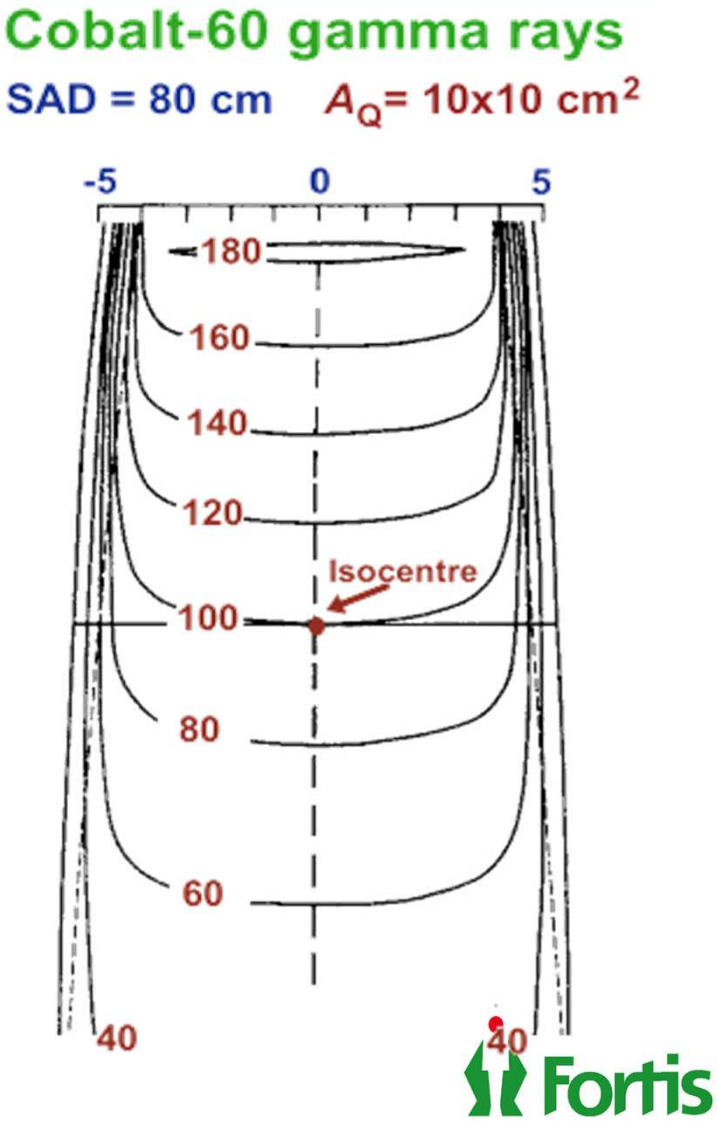
Normalization at isocentre

- For SAD set-ups, the isodose values are normalized to 100% at the isocentre.



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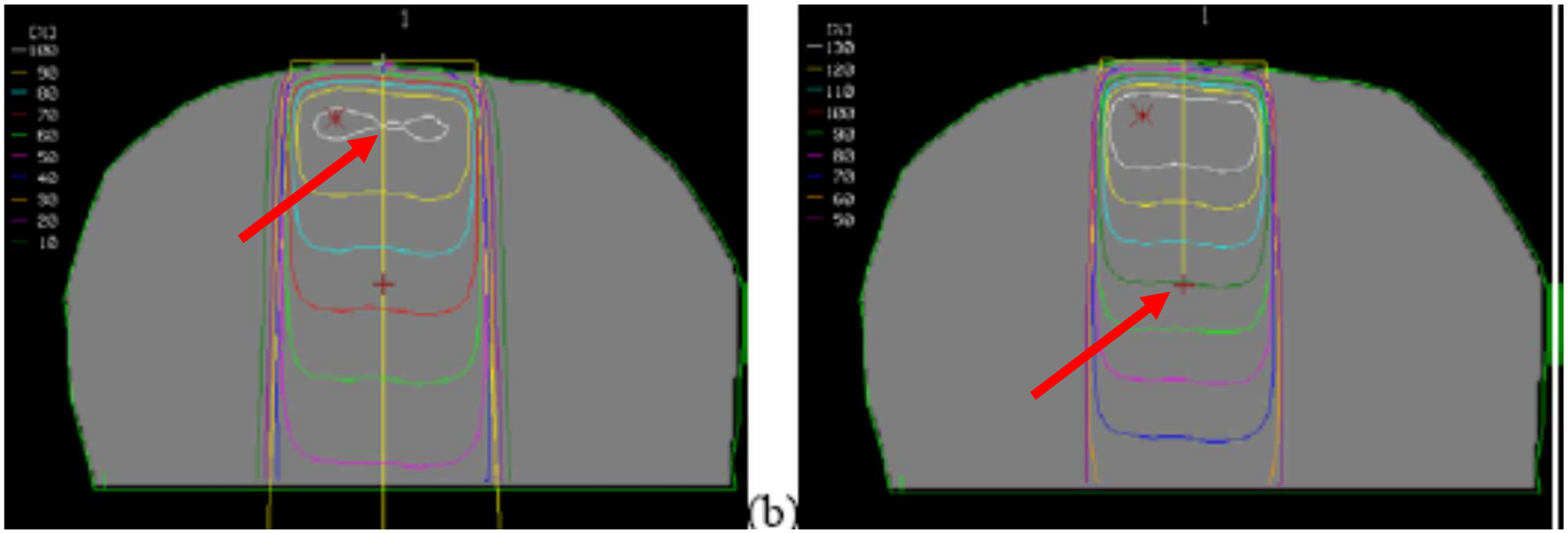
Different type of normalization

Different normalizations for a
single 18 MV photon beam incident on a patient contour



Isodose curves for a fixed SSD beam normalized at depth of **dose maximum**

Isodose curves for an isocentric beam normalized at the **isocenter**

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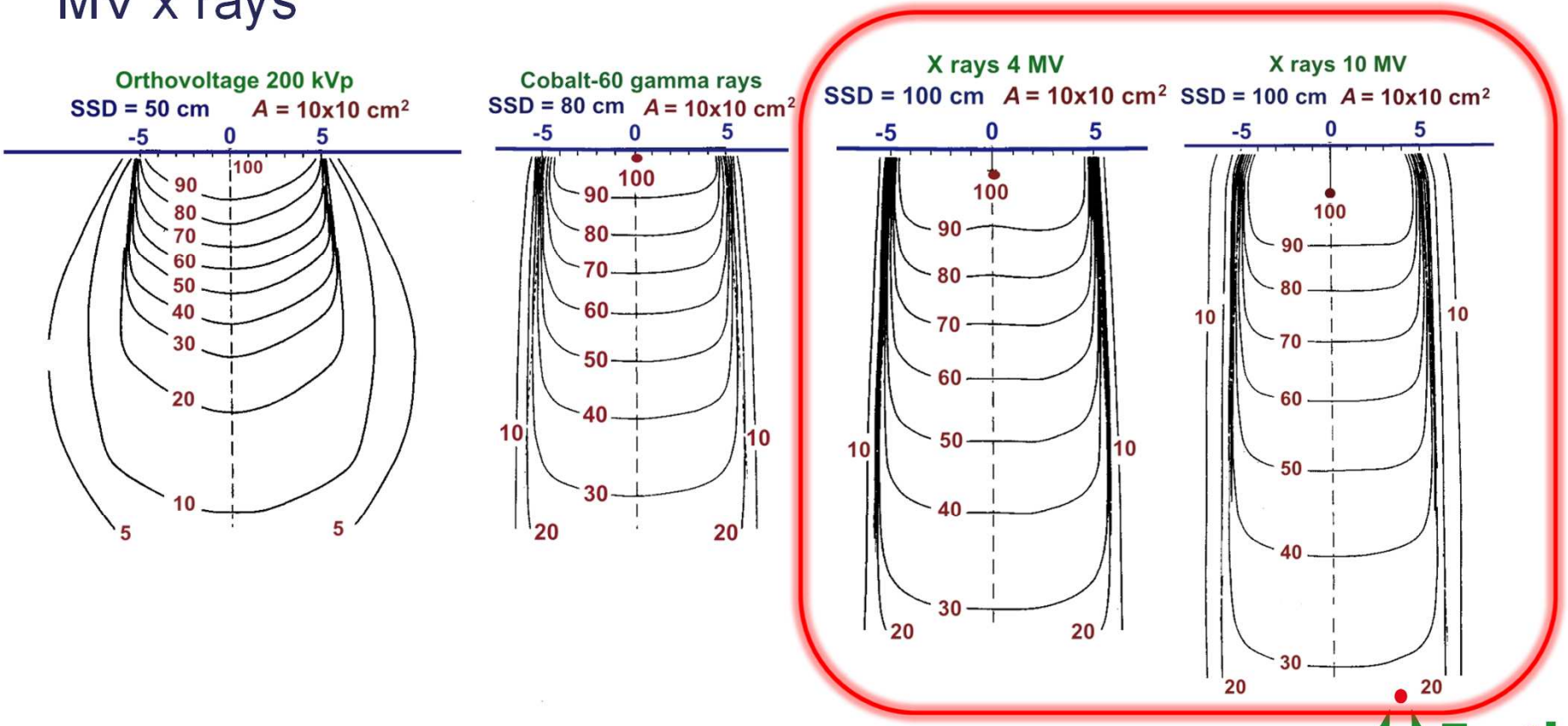
Isodose curves are affected by…

- Beam quality
- Source size
- Beam collimation
- Field size
- Source-skin distance
- Source-collimator distance

Isodose curves for different energies

- Isodose distributions for various photon radiation beams: orthovoltage x rays, cobalt-60 gamma rays, 4 MV x rays, 10 MV x rays



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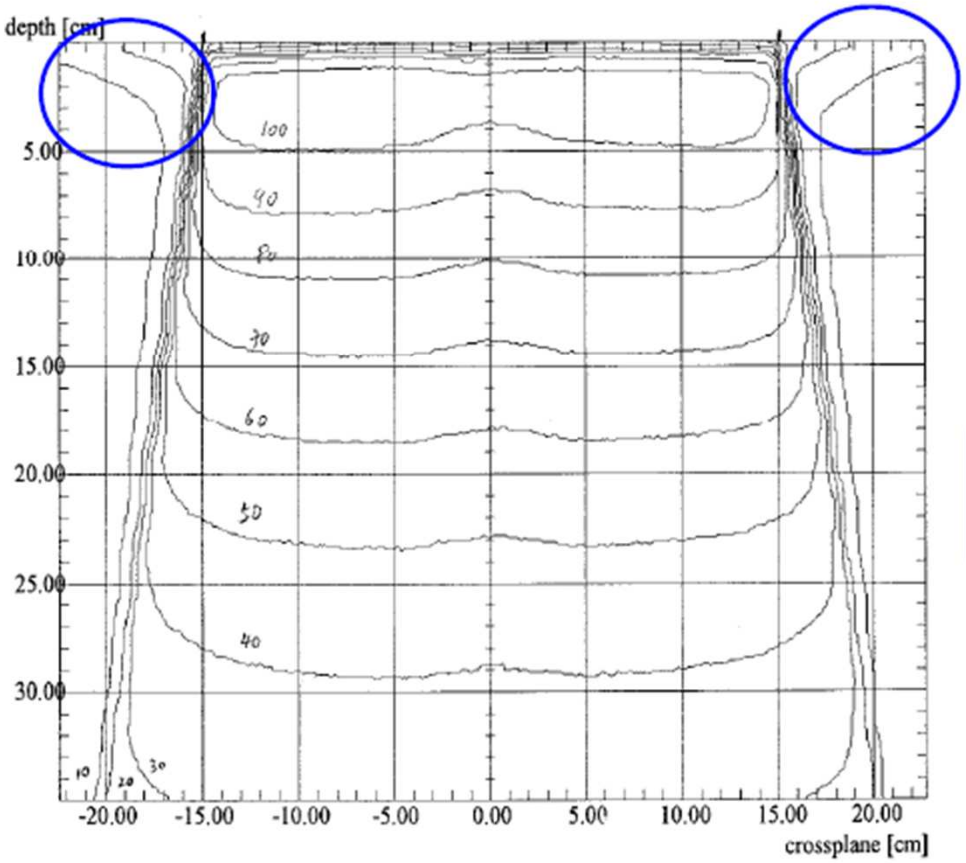
‘Ears’ in an isodose chart: *Have you ever noticed?*

Isodose Distribution



30 cm X 30 cm
18 MV X-ray beam

From: Dr. Palta, Univ of Florida

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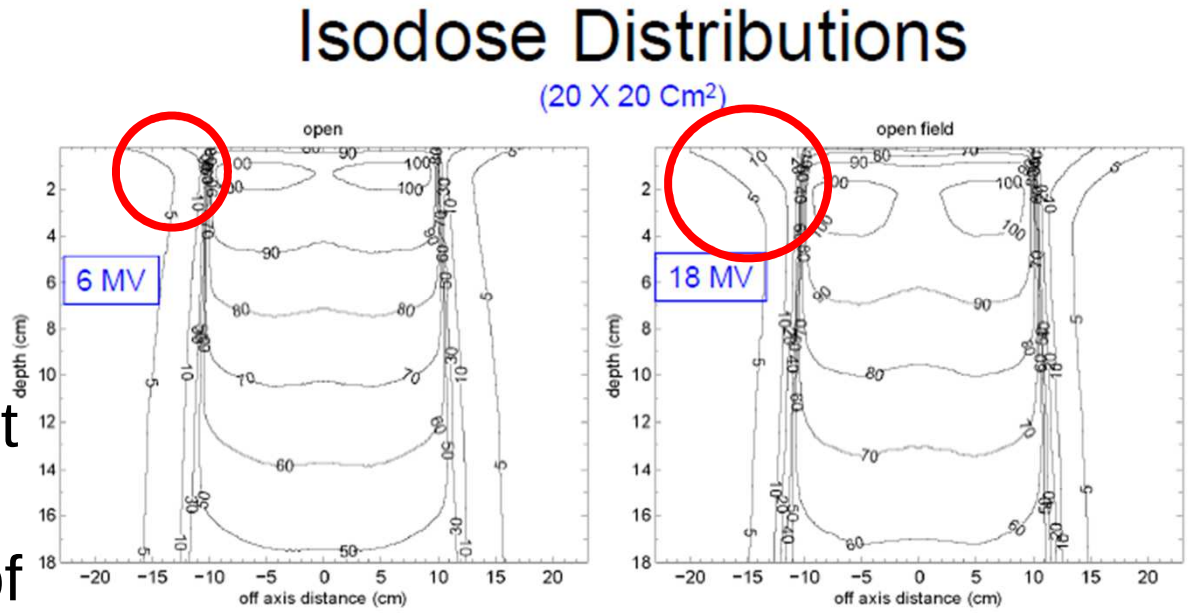
‘Ears’ in an isodose chart: *Have you ever noticed?*

- **Contaminant electrons** contribute to dose outside the field at shallow depths. The magnitude and extent of the dose outside the geometric edge of a field at shallow depths **increases** with beam energy



Does your TPS model this phenomenon?

From: Dr. Palta, Univ of Florida

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Corrections for contour irregularities

- Measured dose distributions apply to a **flat radiation beam** incident on a flat homogeneous water phantom.
- To relate such measurements to the actual dose distribution in a patient, **corrections** for irregular surface and tissue inhomogeneities have to be applied.
- Three methods for contour correction are used:

  (1) the (manual) isodose shift method;

  (2) the effective attenuation coefficient method;

  (3) the TAR method.

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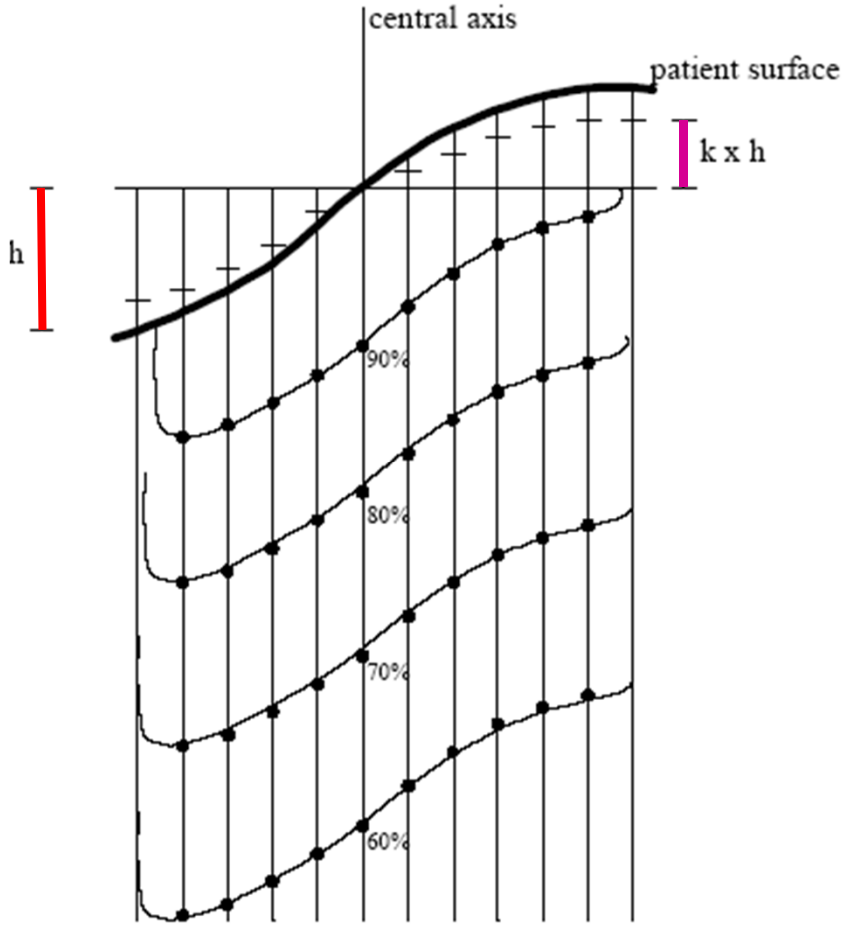
Corrections for contour irregularities

(1) Manual isodose shift method

- Grid lines are drawn parallel to the central beam axis all across the field.
- The tissue deficit (or excess) **h** is the difference between the SSD along a gridline and the SSD on the central axis.
- k is an energy dependent parameter given in the next slide.
- The isodose distribution for a flat phantom is aligned with the SSD central axis on the patient contour.
- For each gridline, the overlaid isodose distribution is shifted up (or down) such that the overlaid SSD is at a point **k×h** above (or below) the central axis SSD.



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Corrections for contour irregularities

Parameter k used in the isodose shift method

Photon energy (MV)	k (approximate)
< 1	0.8
^{60}Co - 5	0.7
5 – 15	0.6
15 – 30	0.5
> 30	0.4

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Corrections for contour irregularities

(2) Effective attenuation coefficient method

- The correction factor is determined from the attenuation factor exp(-μx), where x is the depth of missing tissue above the calculation point, and μ is the linear attenuation coefficient of tissue for a given energy.

- For simplicity the factors are usually pre-calculated and supplied in graphical or tabular form.

Corrections for contour irregularities

(3) TAR method

- The tissue-air ratio (TAR) correction method is also based on the attenuation law, but takes the depth of the calculation point and the field size into account.

- Generally, the correction factor C_F as a function of depth z, thickness of missing tissue h, and field size f, is given by:

$$C_F = \frac{TAR(z-h,f)}{TAR(z,f)}$$

- TMRs / TPRs also can be used in place of TAR

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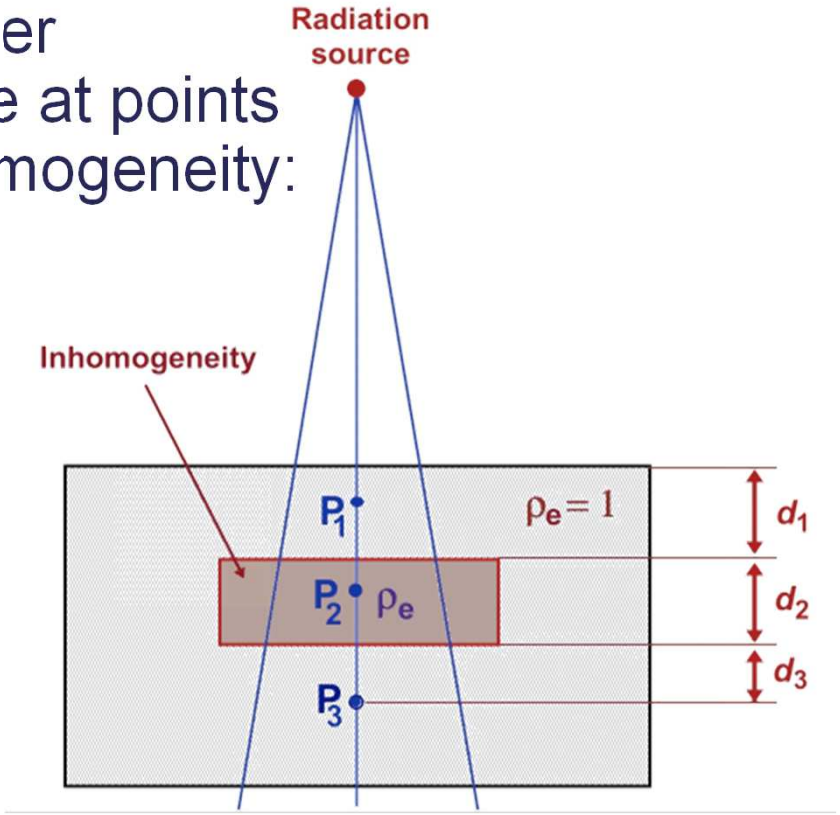
Corrections for tissue inhomogeneities

- Radiation beams used in patient treatment traverse various tissues that may differ from water in density and atomic number.
- This may result in isodose distributions that differ significantly from those obtained with water phantoms.
- The effects of inhomogeneities on the dose distributions depend upon:
  - Amount, density and atomic number of the inhomogeneity.
  - Quality of the radiation beam.

Corrections for tissue inhomogeneities

- Four empirical methods have been developed for correcting the water phantom dose to obtain the dose at points P_3 in region (3) beyond the inhomogeneity:
  - TAR method
  - Power law TAR method
  - Equivalent TAR method
  - Isodose shift method



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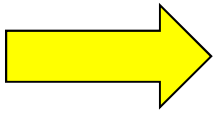
Best way to account for inhomogeneities

- Model based algorithms
  - Convolution-superposition method
  - Monte Carlo method

General considerations for photon beams

- Almost a dogma in external beam radiotherapy:

Successful radiotherapy requires a uniform dose distribution within the target (tumor).



External photon beam radiotherapy is usually carried out with **multiple radiation beams** in order to achieve a **uniform dose distribution** inside the target volume and a dose as low as possible in healthy tissues surrounding the target.

Criteria of a uniform dose distribution within the target

- Recommendations regarding dose uniformity, prescribing, recording, and reporting photon beam therapy are set forth by the International Commission on Radiation Units and Measurements (ICRU).

- The ICRU report 50 recommends a target dose uniformity **within +7% and –5%** relative to the dose delivered to a well defined prescription point within the target.

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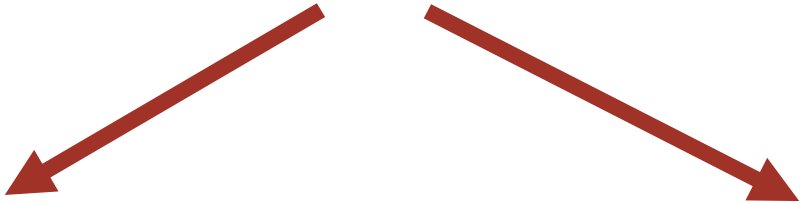
Methods of beam setup

- Photon beam radiotherapy is carried out under two setup conventions



constant Source-Surface Distance

(SSD technique)

isocentric setup with a constant Source-Axis Distance

(SAD technique).

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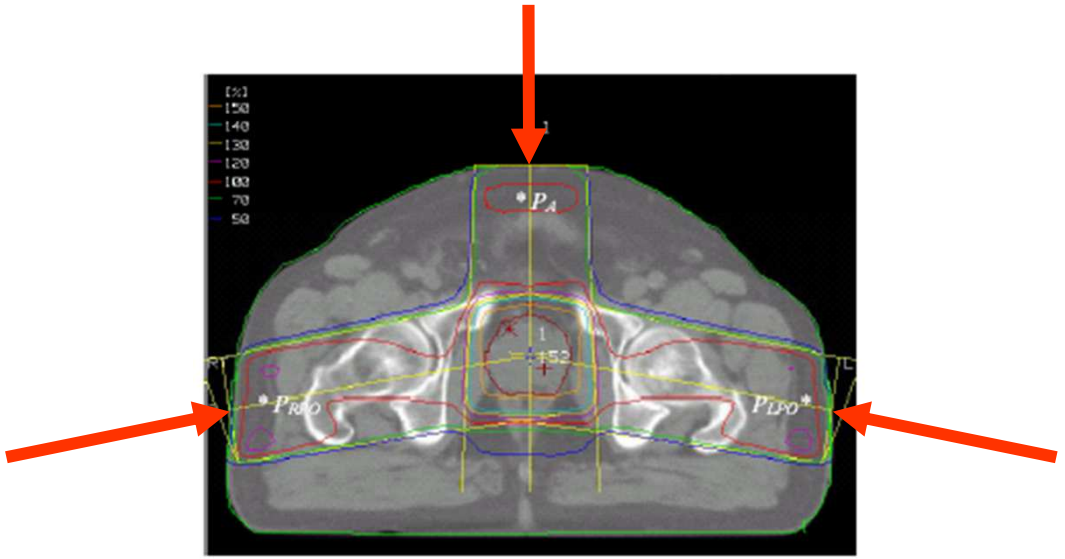
SSD technique

- The distance from the source to the **surface** of the patient is kept **constant** for all beams.



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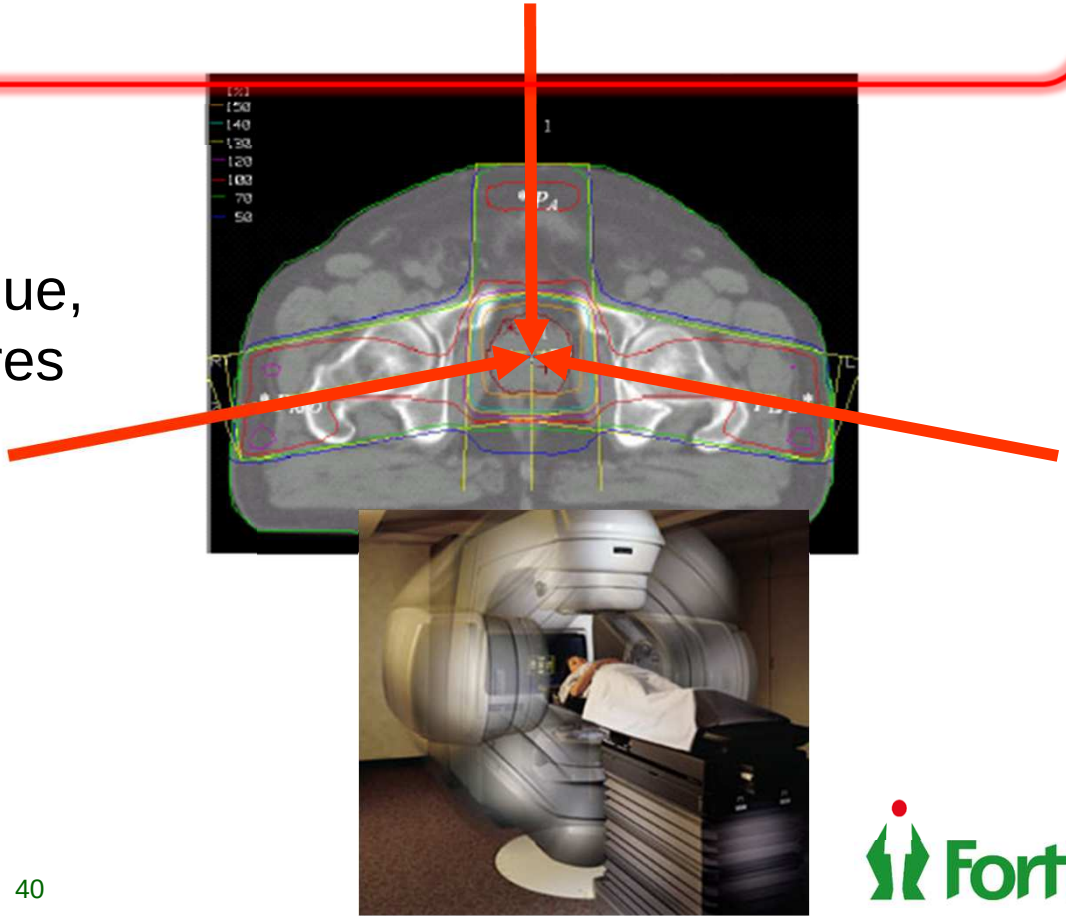
SAD technique

- The center of the target volume is placed at the machine isocenter, i.e. the **distance to the target point** is kept constant for all beams.

Note:

In contrast to SSD technique, the **SAD technique** requires **no** adjustment of the patient setup when turning the gantry to the next field.



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SSD vs. SAD technique: *Which is better?*

- There is little difference between fixed SSD techniques and isocentric (SAD) techniques with respect to the dose:
  - Fixed SSD arrangements are usually at **a greater SSD** than isocentric beams because the machine isocenter is on the patient skin.
  - They have therefore a slightly **higher PDD** at depth.
  - Additionally, beam **divergence is smaller** with SSD due to the larger distance.

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SSD vs. SAD technique: *Which is better?*

- These dosimetric advantages of SSD technique are small.
- With the exception of very large fields exceeding 40x40 cm^2, **the advantages of using a single set-up point** (i.e., the isocenter) greatly outweigh the dosimetric advantage of SSD beams

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Dose specification

- Parameters to characterize the dose distribution within a volume and to specify the dose are:
  - Minimum target dose
  - Maximum target dose
  - Mean target dose
  - A **reference dose** at a **representative point** within the volume
- The ICRU has given recommendations for the selection of a representative point (the so-called **ICRU reference point**).

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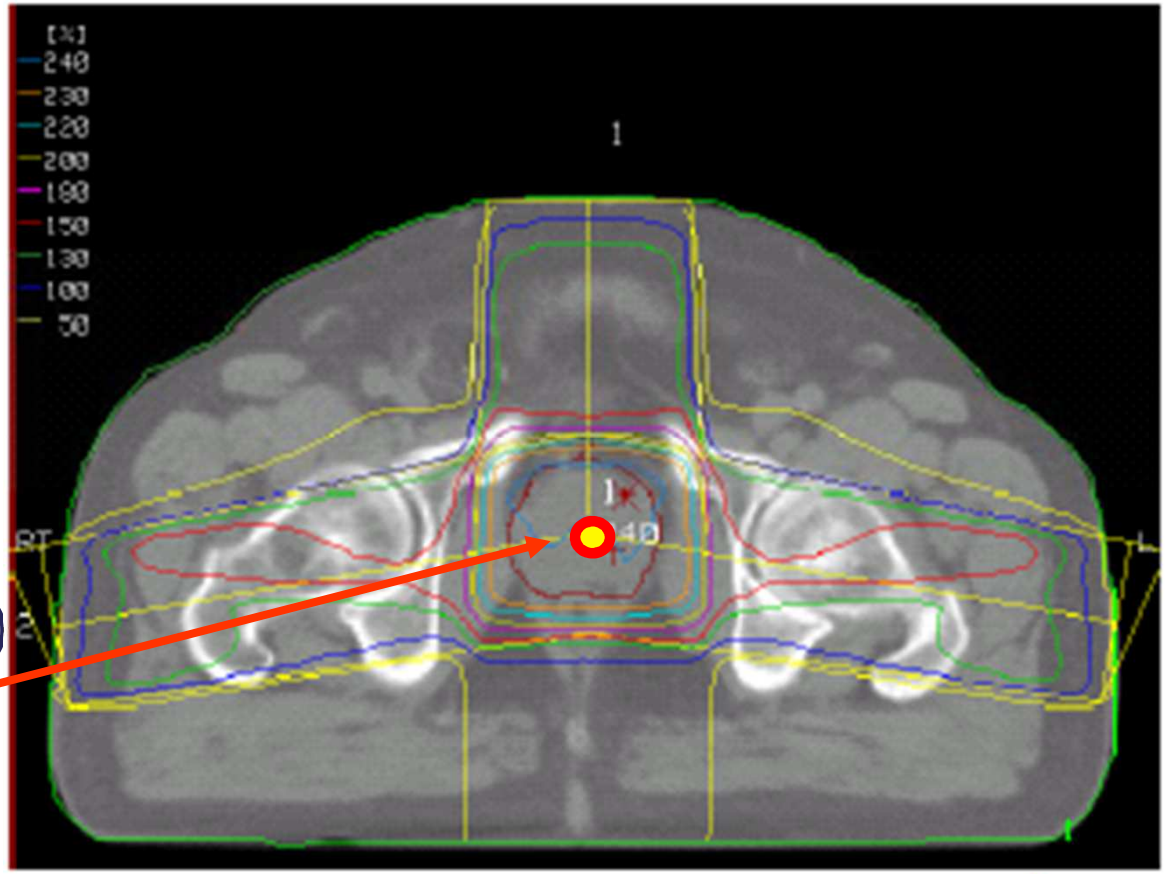
Dose specification

- The **ICRU reference dose point** is located at a point chosen to represent the delivered dose using the following criteria:
  - The point should be located in a region where the dose can be calculated accurately (i.e., no build-up or steep gradients).
  - The point should be in the central part of the PTV.
  - For multiple fields, the isocenter (or beam intersection point) is recommended as the ICRU reference point.

Dose specification

ICRU reference point for multiple fields

Example for a 3 field prostate boost



The ICRU (reference) point is located at the isocenter

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ICRU Reference point

- Specific recommendations are made with regard to the position of the ICRU (reference) point for particular beam combinations:
  - **For single beam:**
    the point on **central axis** at the **center of the target volume**.
  - **For parallel-opposed equally weighted beams:**
    the point on the **central axis midway between the beam entrance points**.
  - **For parallel-opposed unequally weighted beams:**
    the point on the central axis at the **centre of the target volume**.
  - **For other combinations of intersecting beams:**
    the **point at the intersection of the central axes** (insofar as there is no dose gradient at this point).

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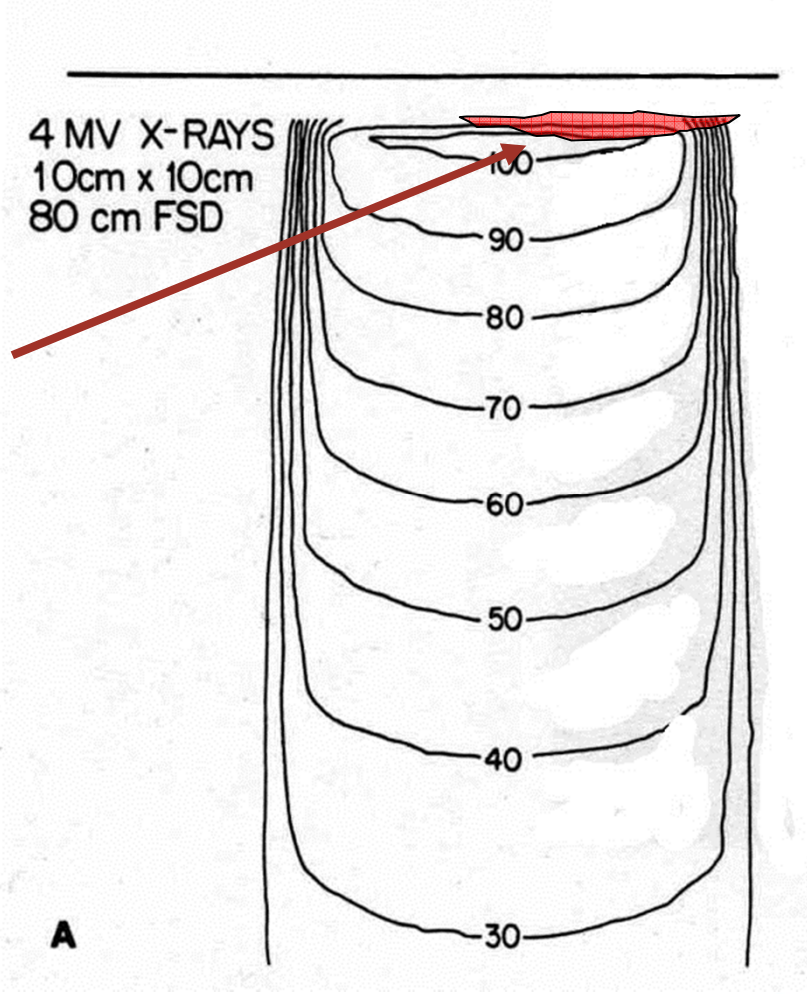
Beam combinations and clinical application

- Single photon beams are of limited use in the treatment of deep-seated tumors, since they give a **higher dose** near the entrance at the depth of dose maximum than at depth.



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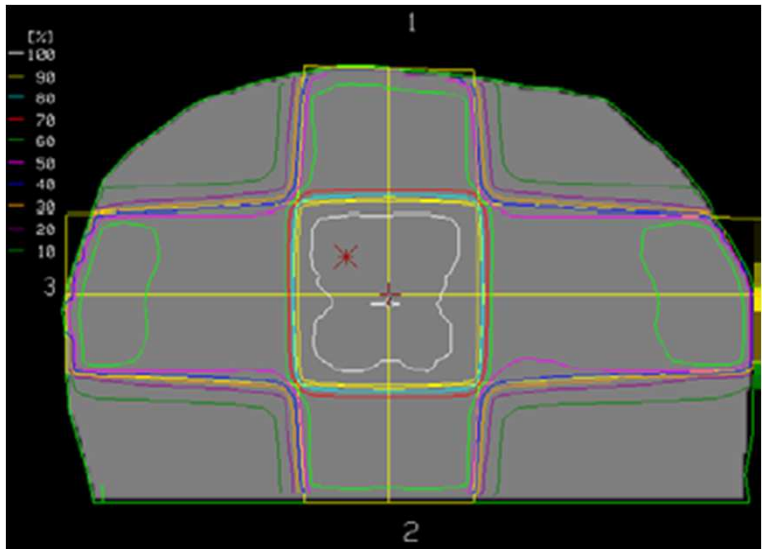
Beam combinations and clinical application

- Single fields are often used for **palliative treatments** or for relatively **superficial lesions** (depth < 5-10 cm, depending on the beam energy).

- For deeper lesions, a **combination** of two or more photon beams is usually required to concentrate the dose in the target volume and spare the tissues surrounding the target as much as possible.



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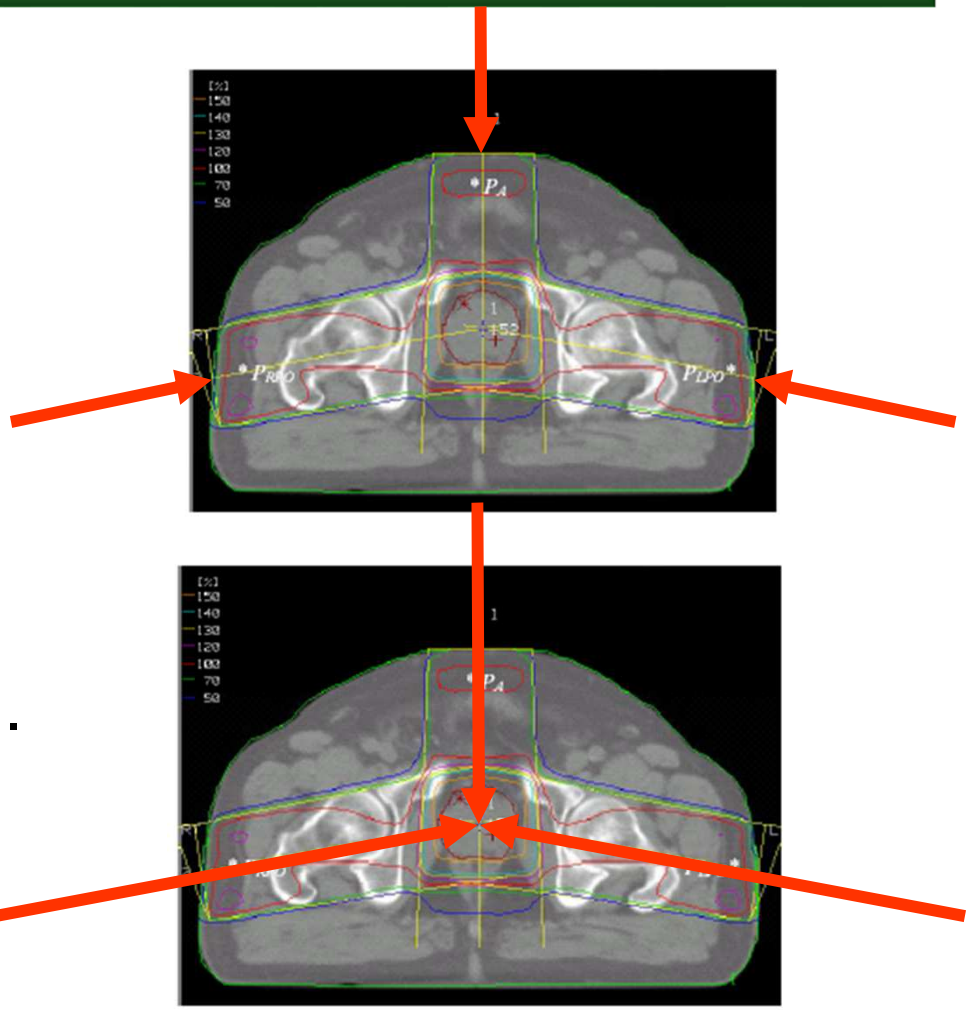
Beam combinations and clinical application

Normalization

- Dose distributions for multiple beams can be normalized to 100% just as for single beams:
  - at z_{max} for each beam,
  - at isocenter for each beam.
- This implies that each beam is equally weighted.



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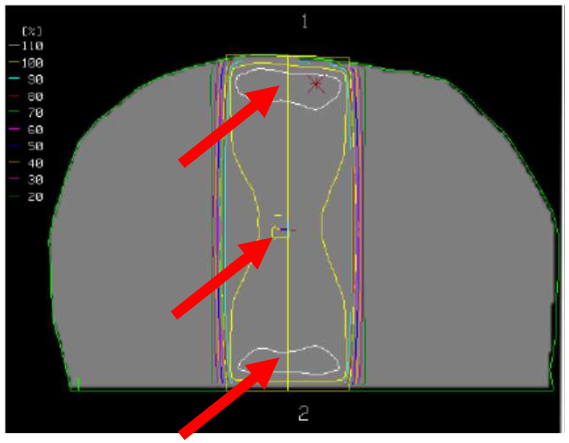
Beam combinations and clinical application

Weighting and normalization

- A **beam weighting** may additionally applied at the normalization point for the given beam.



- Example:

  Two beams with z_{max} normalization weighted as

  100 : 50%

  will show one beam with the 100% isodose at z_{max} and the other one with 50% at z_{max}.

- A similar isocentric weighted beam pair would show the 150% isodose at the isocenter.

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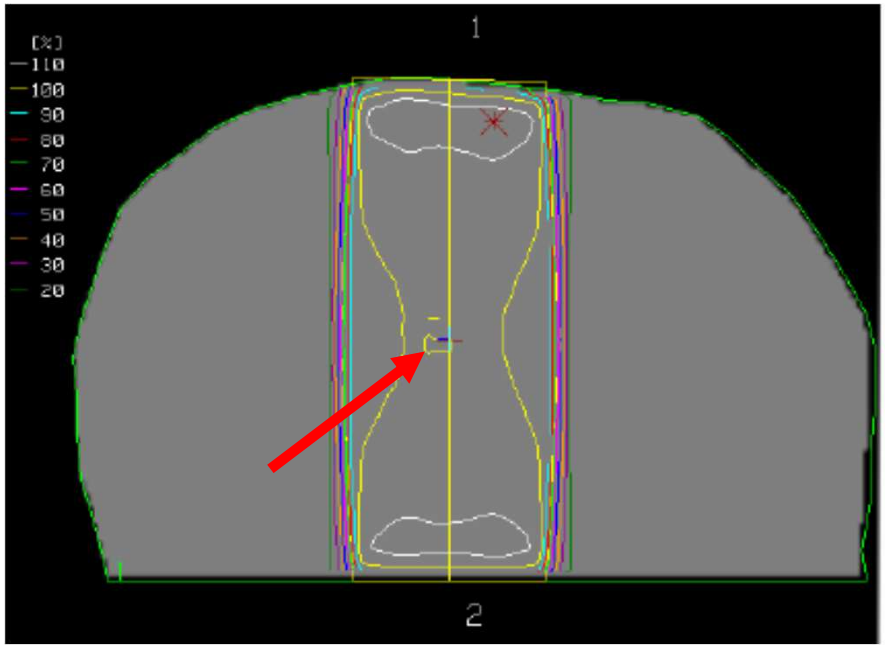
Beam combinations and clinical application

Parallel opposed beams – Equally weighted

- Example:
  A parallel-opposed beam pair is incident on a patient.
- Note the large rectangular area of relatively uniform dose (<15% variation).
- The isodoses have been normalized to 100% at the isocenter.



- This beam combination is well suited to a large variety of treatment sites (e.g., lung, brain, head and neck).

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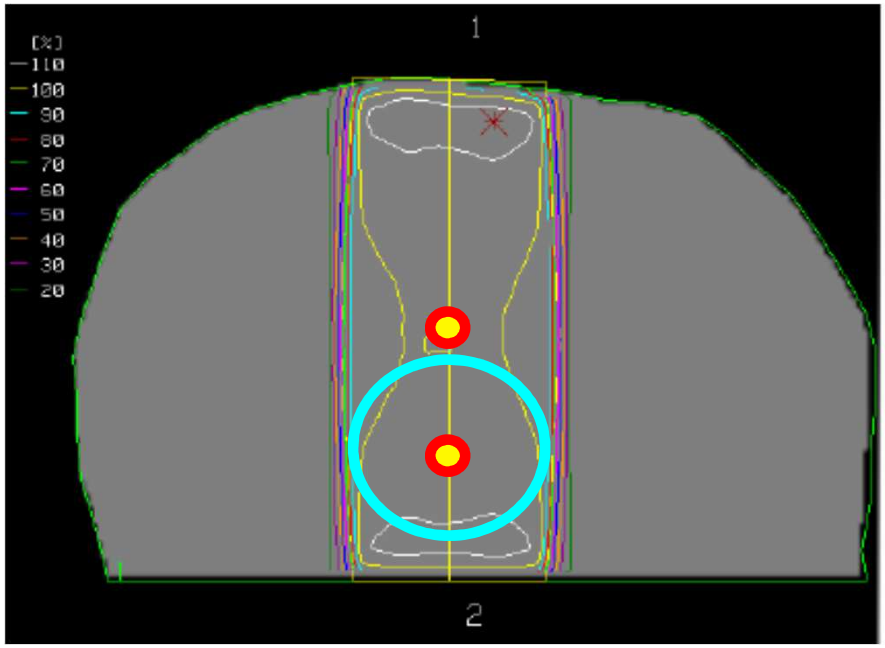
Beam combinations and clinical application

Parallel opposed beams – Unequally weighted

- When?
- Target volume is one sided, but at a larger depth
  - Single beam will give very high entry dose
  - Equally weighted opposing beams will give high dose throughout the volume



- Give more weight to the beam from target side & less to the other
- Remember, it is **NOT** a magic solution
- Isocenter can be at mid-plane or at center of tumor – ratio of weights will differ

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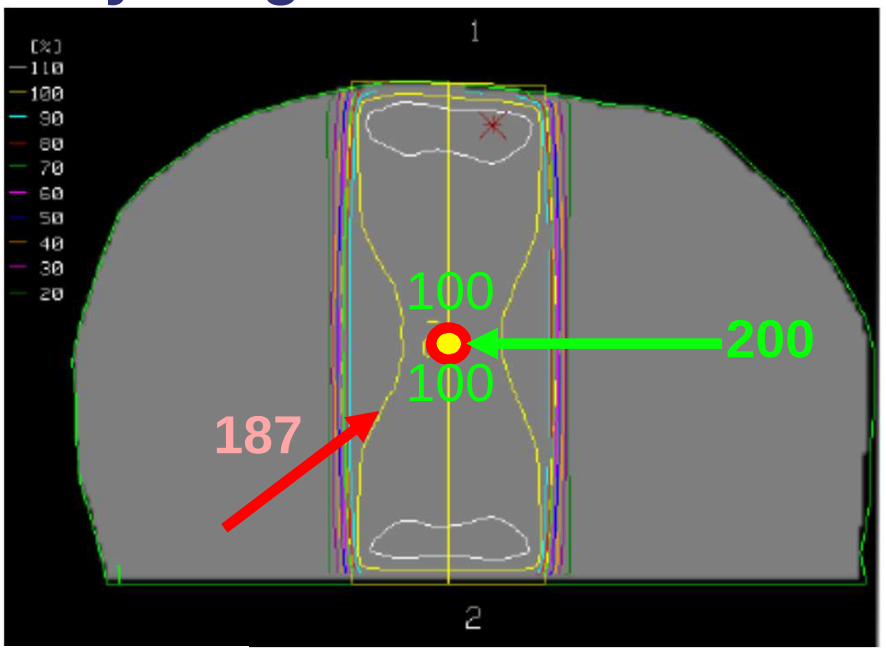
Beam combinations and clinical application

Parallel opposed beams – Equally weighted

- Weight - 100:100 at iso
- At iso, it is now 200%
- Tumor is covered by 187%
- What does this mean?



	Beam 1	Beam 2
Tumor 187 cGy	100 cGy	100 cGy
Tumor 300 cGy	$\left(\frac{100\ cGy}{187\ cGy}\right) * 300\ cGy = 160.4\ cGy$	

We need to deliver this 160.4 cGy **at the iso of each beam**

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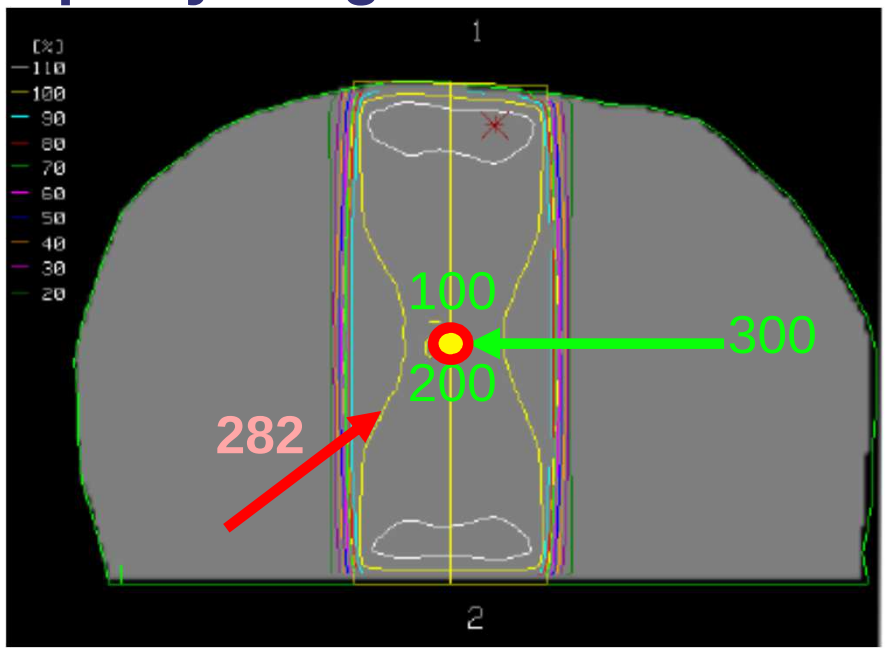
Beam combinations and clinical application

Parallel opposed beams – Unequally weighted

- Weight - 100:200 at iso (AP:**PA**)
- At iso, it is now 300%
- Tumor is covered by 282%
- What does this mean?



	Beam 1	Beam 2
Tumor 282 cGy	100 cGy	200 cGy
Tumor 300 cGy	$\left(\frac{100\ cGy}{282\ cGy}\right) * 300\ cGy = 106.4\ cGy$	$\left(\frac{200\ cGy}{282\ cGy}\right) * 300\ cGy = 212.8\ cGy$

We need to deliver 106.4 cGy & 212.8 cGy **at the iso of AP & PA beams respectively**

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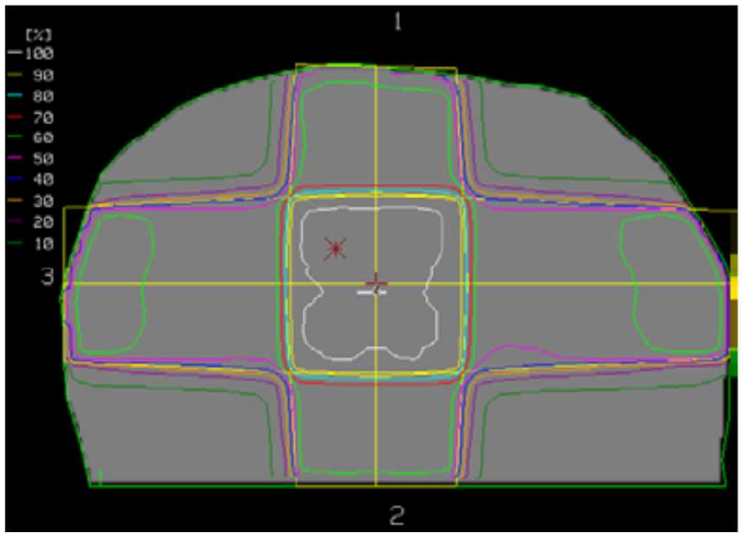
Beam combinations and clinical application

Multiple co-planar beams

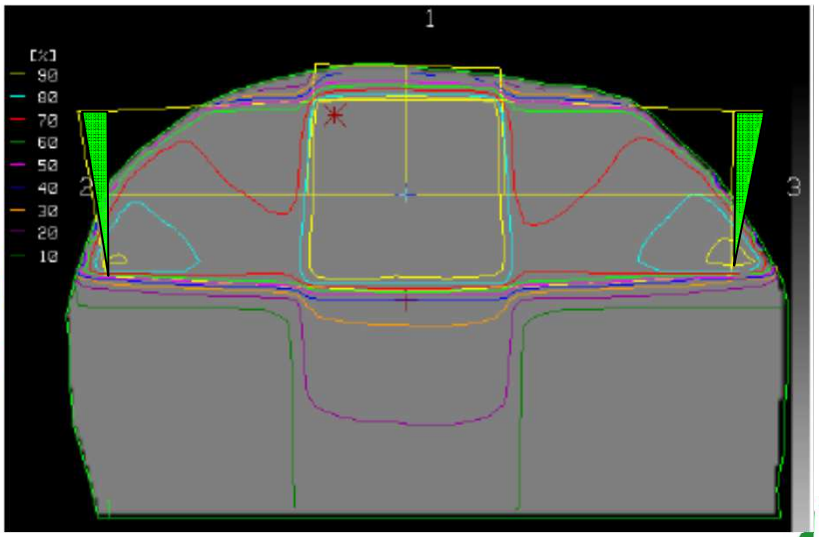
- Multiple coplanar beams allows for a higher dose in the beam intersection region.

Two examples:

4-field box



3-field technique using wedges



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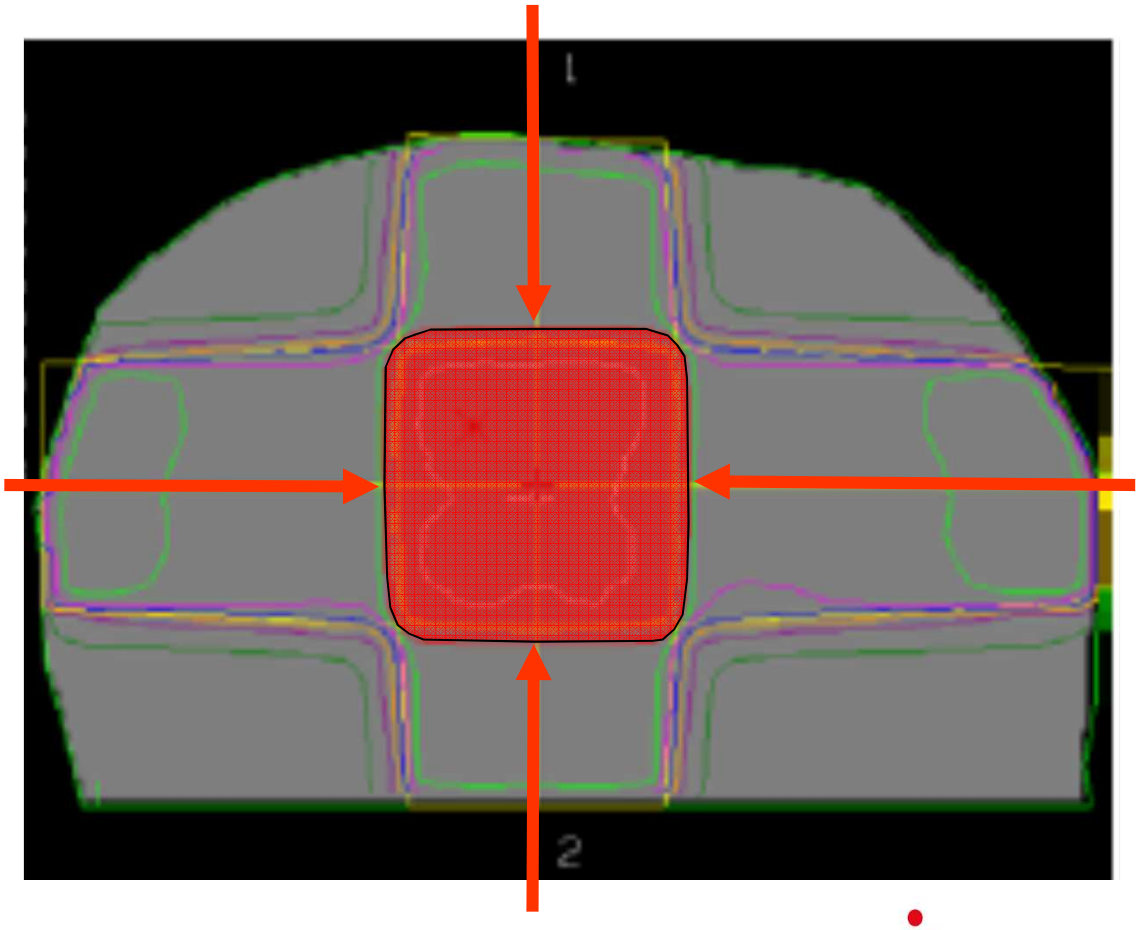
Beam combinations and clinical application

Multiple co-planar beams – 4 field box

- A 4-field box allows for a very high dose to be delivered at the **intersection** of the beams.



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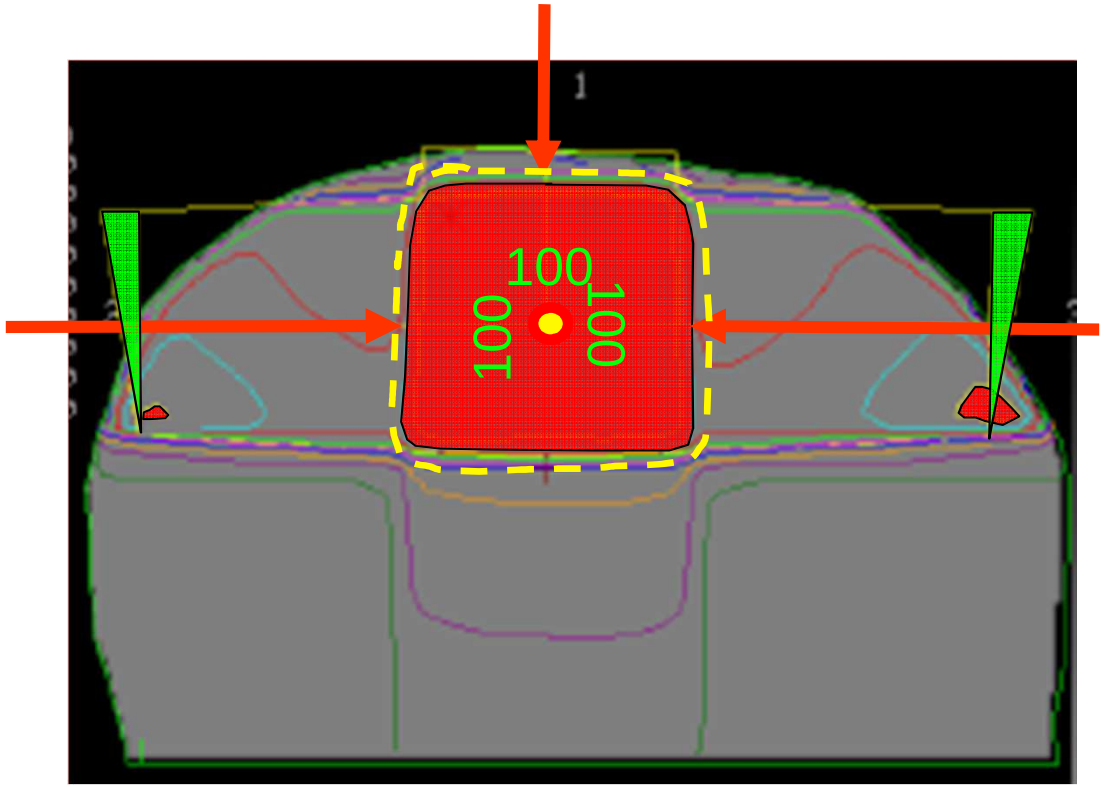
Beam combinations and clinical application

Multiple co-planar beams – 3-field technique using wedges

- A 3-field technique requires the use of wedges to achieve a **similar result**.

- Note that the latter can produce significant **hot spots** near the entrance of the wedged beams and well outside the targeted area



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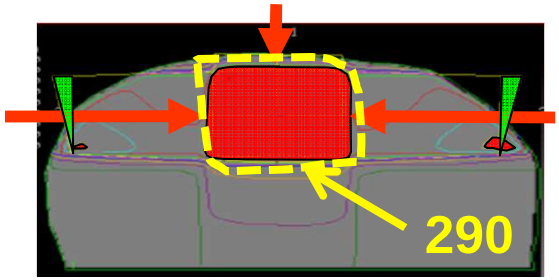
Beam combinations and clinical application

3-field technique using wedges – *Equally weighted*

- Weight 100:100:100 at iso
- At iso it is now 300%
- Target covered by 290%



	Beam 1	Beam 2	Beam 3
Target 290 cGy	100 cGy	100 cGy	100 cGy
Target 200 cGy	$\left(\frac{100\ cGy}{290\ cGy}\right) * 200\ cGy = 69\ cGy$		

We have to deliver 69 cGy at the iso of each beam

Note: For the wedged beams, we need to take into account the wedge attenuation factor (transmission factor) while calculating time or MUs

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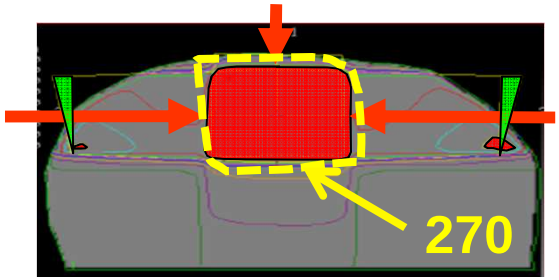
Beam combinations and clinical application

3-field technique using wedges – *Unequally weighted*

- Weight 60:110:120 at iso (AP:RL:LL)
- At iso it is now 290%
- Target covered by 270%



	Beam 1	**Beam 2**	**Beam 3**
Target 270 cGy	60 cGy	110 cGy	120 cGy
Target 200 cGy	$\left(\frac{60\ cGy}{270\ cGy}\right) * 200\ cGy = 44.4\ cGy$	$\left(\frac{110\ cGy}{270\ cGy}\right) * 200\ cGy = 81.5\ cGy$	$\left(\frac{120\ cGy}{270\ cGy}\right) * 200\ cGy = 88.9\ cGy$

We have to deliver 44.4 cGy, 81.5 cGy & 88.9 cGy at the iso of AP, RL and LL beams respectively.

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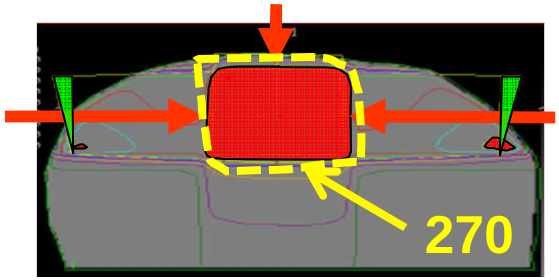
Normalization

- Weight 60:110:120 at iso (AP:RL:LL)
- At iso it is now 290%
- Target covered by 270%
- **We normalize 290% to 100%**



- Target is covered by 270% or [100/290]*[270] = 93.1%
- The question is: are you going to prescribe your dose to 100%? or to 93.1%?
  - Prescribing at 100% - tumor receives a minimum dose of 93.1% of your prescription dose
  - Prescribing at 93.1% - tumor receives your FULL prescription dose, but some part is overdosed by 7.4% (or even more)

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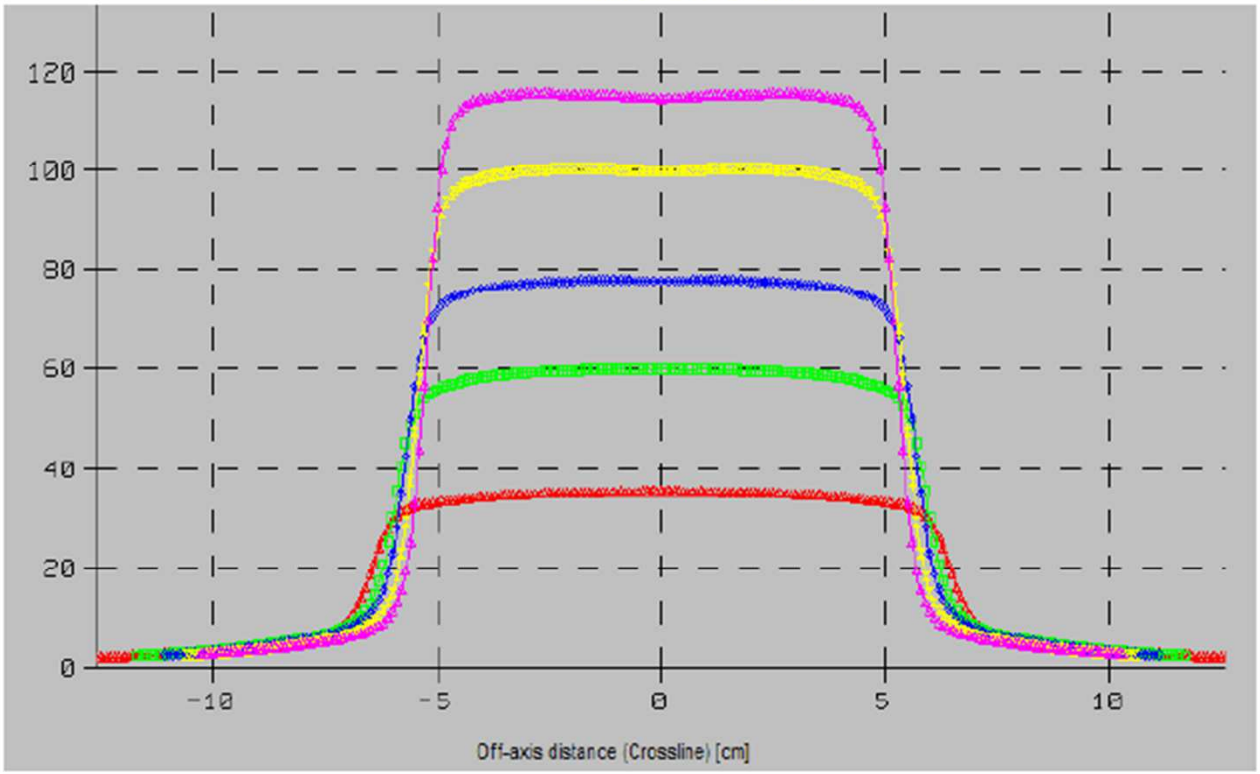
Any questions?

Cross Beam Profile

6 MV Photon Beam, Field Size of 10.4x10.4 cm2, Depths of 1.5, 5.0, 10.0, 15.0, and 25.0 cm.



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

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