PDD Curves
Photons and charged particle beams

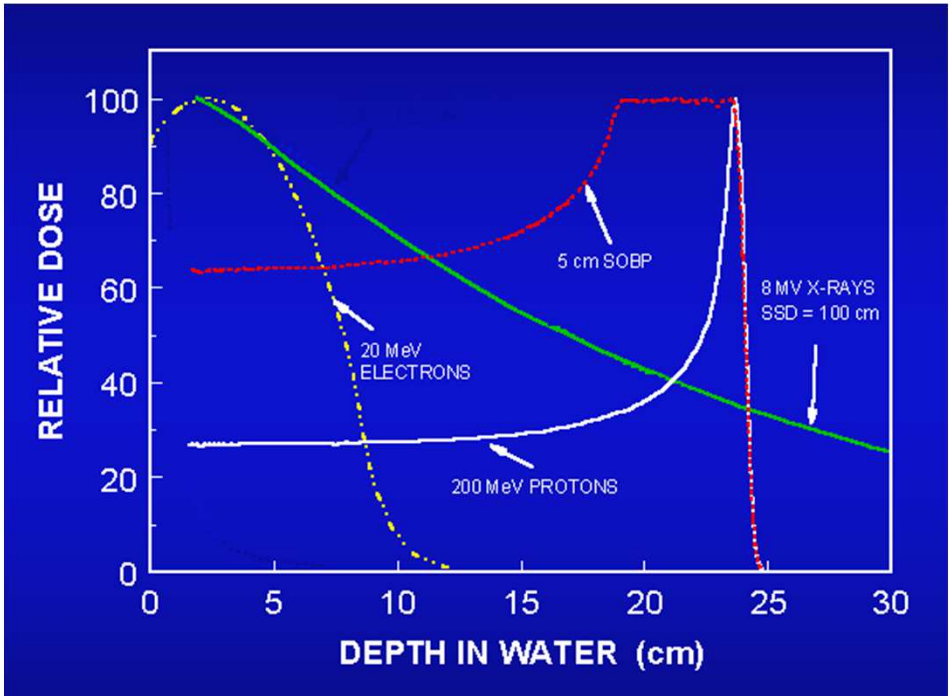
The photon beams differ from particle beams:

Photon beams have high range

Rapid fall-off of dose

Skin sparing effect is negligible electron beams

Photons are charge less particle Electrons have charge of 1.6x10^{-19} c.

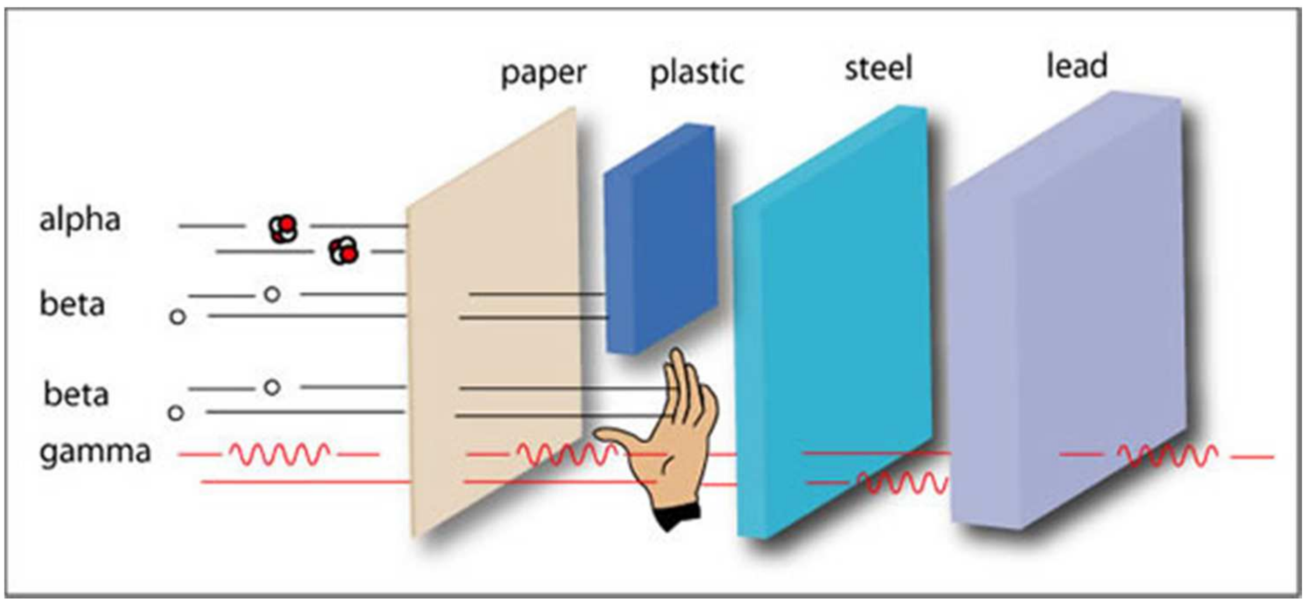


Particle beam can be deflected by magnetic field

Photons vs. Charged Particles

Since photons have no charge, they interact with matter differently than charged particles

For photons, we discuss the probability of interaction per unit distance travelled



Photon beam characteristics & basic concepts of treatment planning

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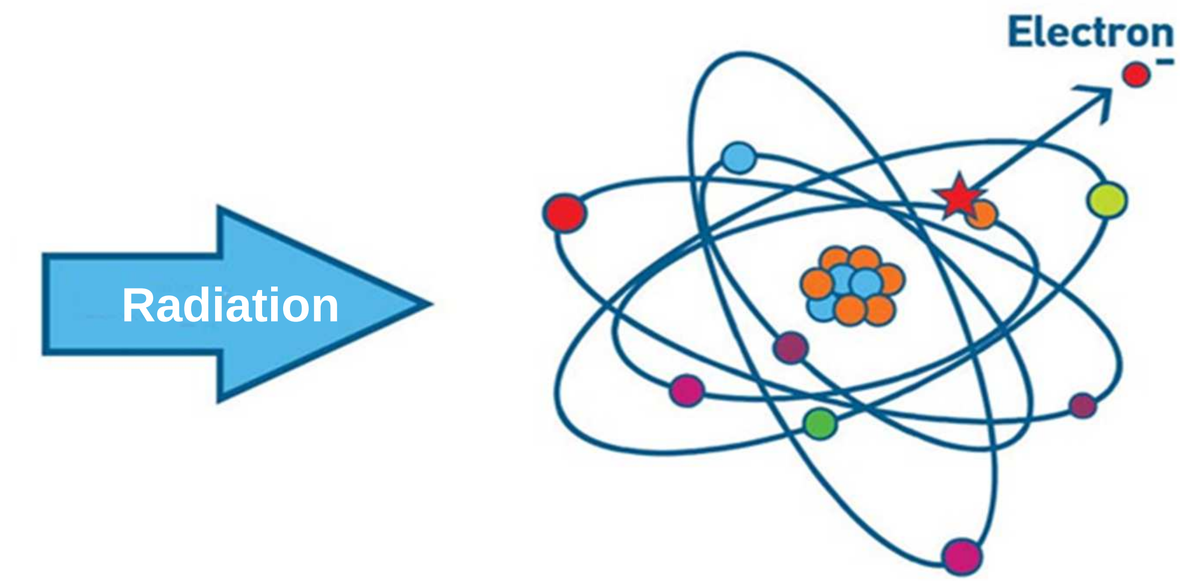
Lucknow

Learning Objectives

- Understanding basic properties of clinical photon beams
- Understanding the parameters that influence the beam profile characteristics
- Influence of beam modifiers
- Dose distribution
- Basic concepts of treatment planning

Ionizing Radiation

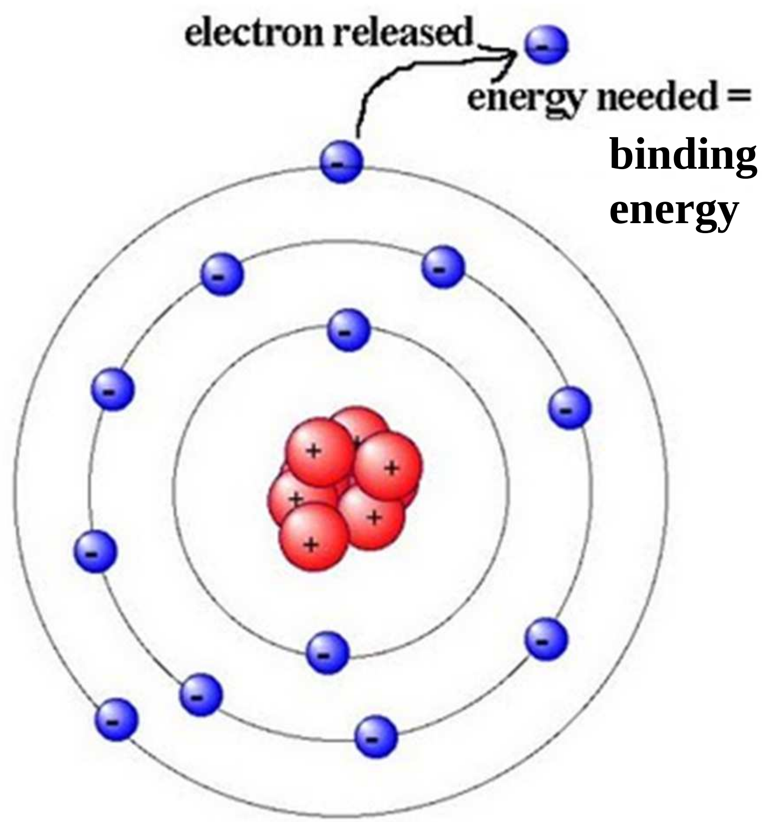


Ionizing radiation has sufficient energy to remove orbital electrons from atoms or molecules with which it interacts.

The specific interaction that occurs depends on the type of radiation.

Electron Binding Energy

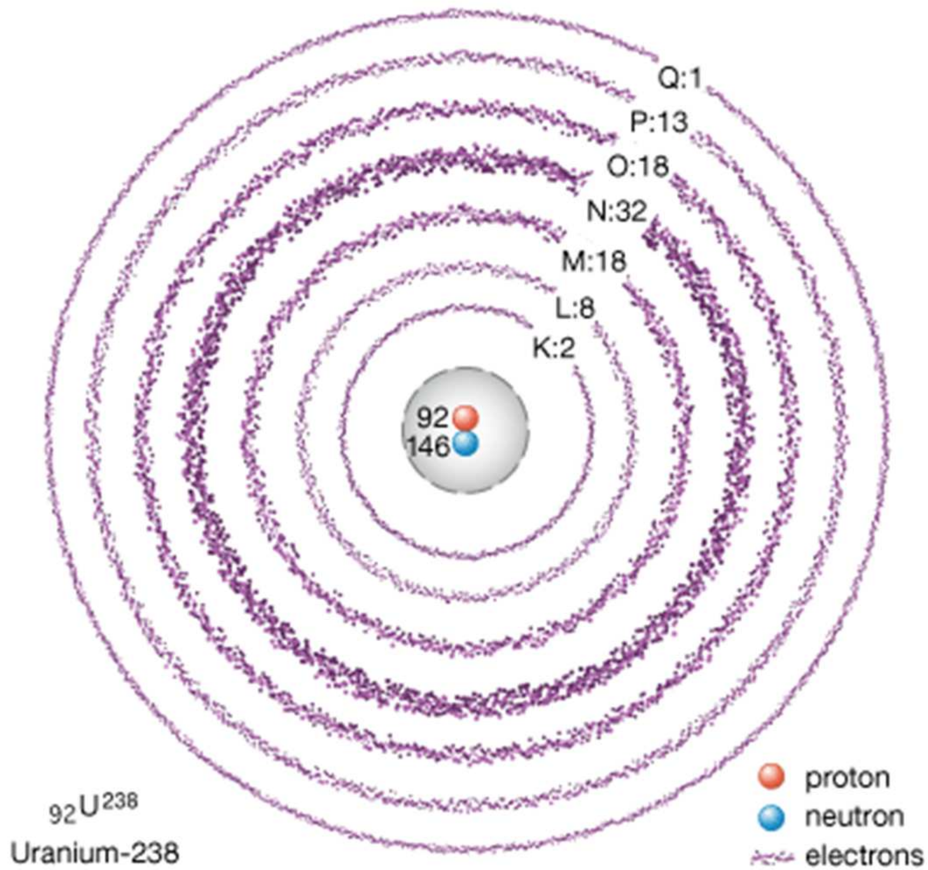
- Binding energy is the energy required to "pull" an electron away from it's positively charged nucleus
- Recall that electrons exist in discrete "shells" around the nucleus. Electrons found in the different shells have different binding energies.



Electron Shells

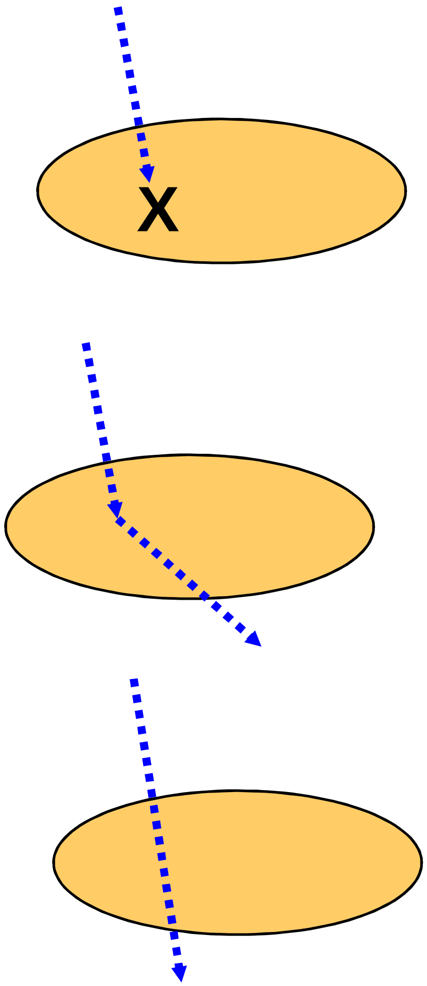
- **The shells are designated by letters (K, L, M, N…) where K, the shell closest to the nucleus, has the largest binding energy, so the K electron is the most tightly bound.**

- **There are a maximum number of electrons in each shell: 2 in K shell, 8 in L shell, etc.**



Photons atoms interactions

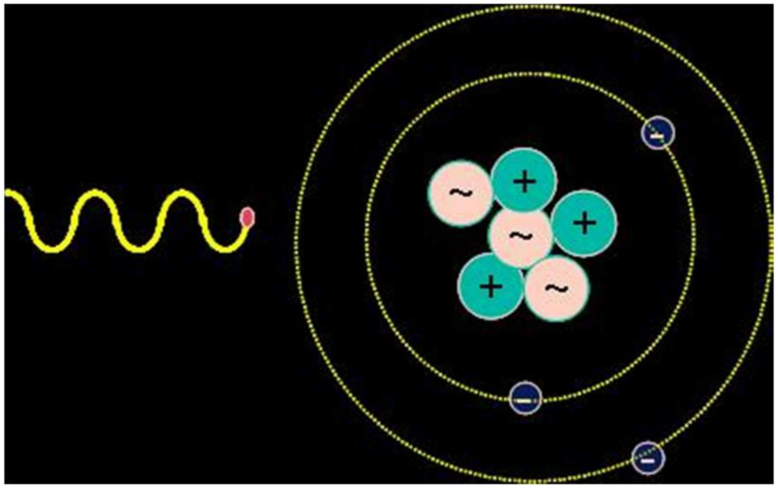
- What happen when photons interact with human tissue?



- **Absorbed**
  - completely removed from beam
  - ceases to exist
- **Scattered**
  - change in direction
  - no useful information carried
  - source of <u>noise</u>
- **Nothing**
  - Photon passes unmolested

Interactions of radiation with matter

- Photon energy $E = h\nu$, 1 eV = $1.602\ 0^{-19}$ J
- Atom atomic number Z
- Electron density e



How the interaction happen?
When happen?
How the interactions affect the beam characteristics?

INTERACTION PROCESS	SYMBOL	INTERACTION TYPE	IMPORTANCE IN RADIOLOGY AND RADIOTHERAPY
Rayleigh scatter	coh	Elastic scattering from atom	Small, but significant at low-energy diagnostic x-rays.
Compton	c	Inelastic scattering from electron	Dominant over wide range of diagnostic and therapy x-rays.
Photo-electric	pe	Atomic absorption	Dominant at low energies and high-Z media.
Pair production	pp	Nuclear absorption	Only above 1.022 MeV. Important in high-energy beams.
Photo-nuclear	pn	Nuclear absorption	Not relevant in radiotherapy or diagnostics.

Coherent or Rayleigh scattering

- **Change in direction**

- **No change in**
  - **energy**
  - **frequency**
  - **Wavelength**

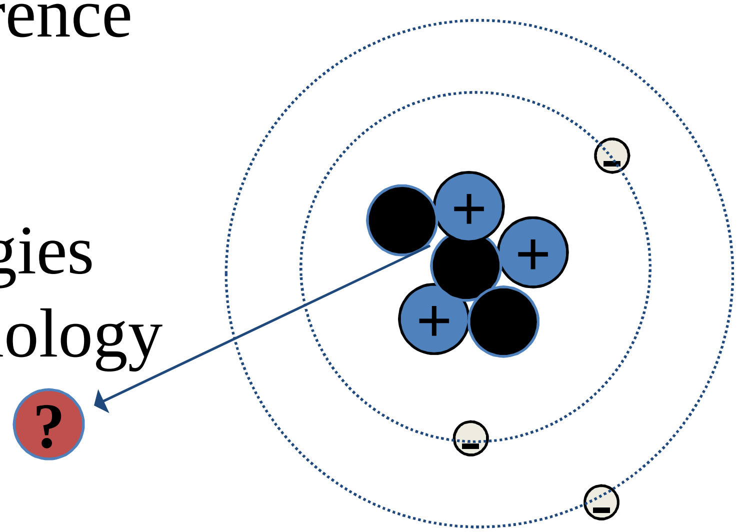
- **No ionization**

- **Scattering probable – High atomic no. materials and with photon of low energy**

- **Less than 5% of interactions insignificant effect on image quality compared to other interactions**

Photodisintegration

- photon causes ejection of part of atomic nucleus
- ejected particle may be
  - neutron
  - proton
  - alpha
  - particle cluster
- Threshold photon energy for occurrence
  - nuclear binding energy
    - typically 7-15 MeV
- Threshold is above diagnostic energies
  - <u>does not occur</u> in diagnostic radiology



Photon Interactions

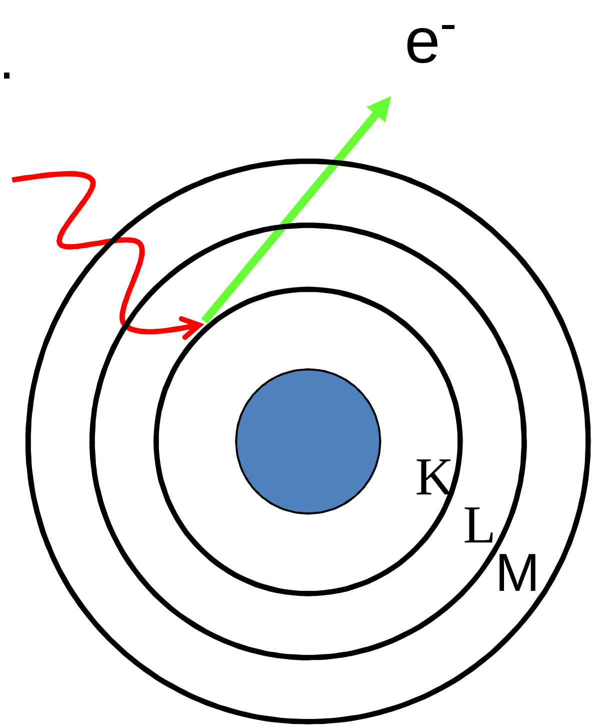
Photons interactions that are important to health physics:

- **Photoelectric Effect**
- **Compton Scattering**
- **Pair Production**

Photoelectric Effect

The photoelectric effect is the predominant interaction mechanism for low energy photons.

1. Incoming photon interacts with an atom as a whole.

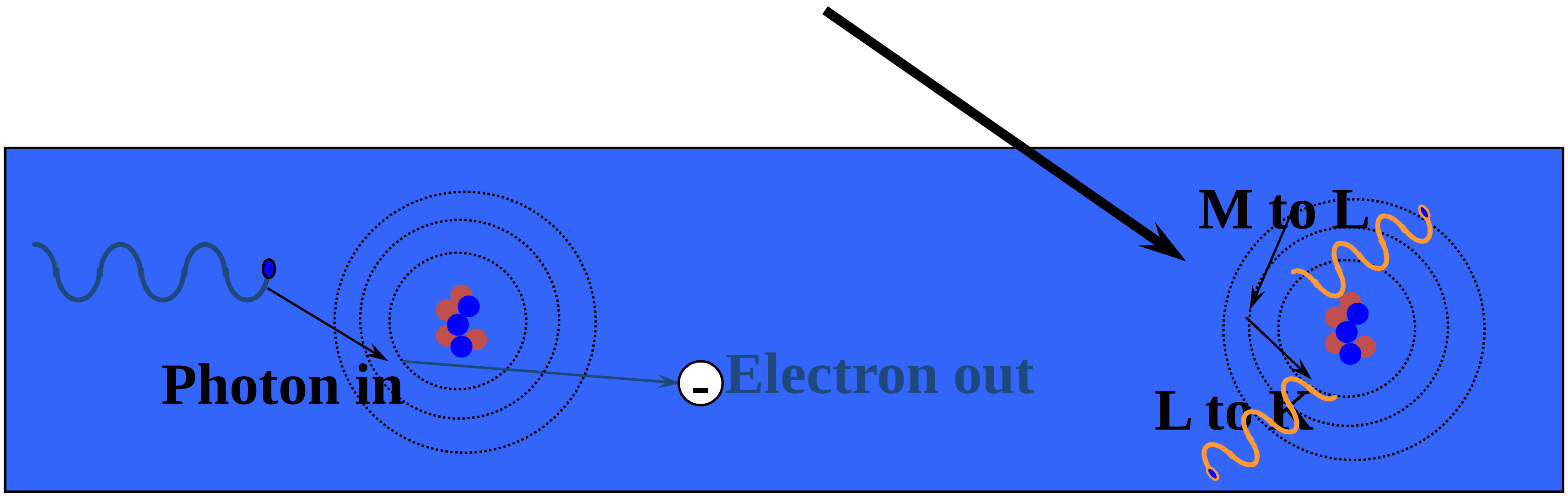


2. Photon disappears after giving up all its energy, and an electron (usually from the K-shell) is ejected from the atom.

Photoelectric Effect

- Exiting electron kinetic energy
  - incident energy - electron's binding energy

- Electrons in higher energy shells cascade down to fill energy void of inner shell
    - **characteristic radiation**



Photoelectric Interaction Probability

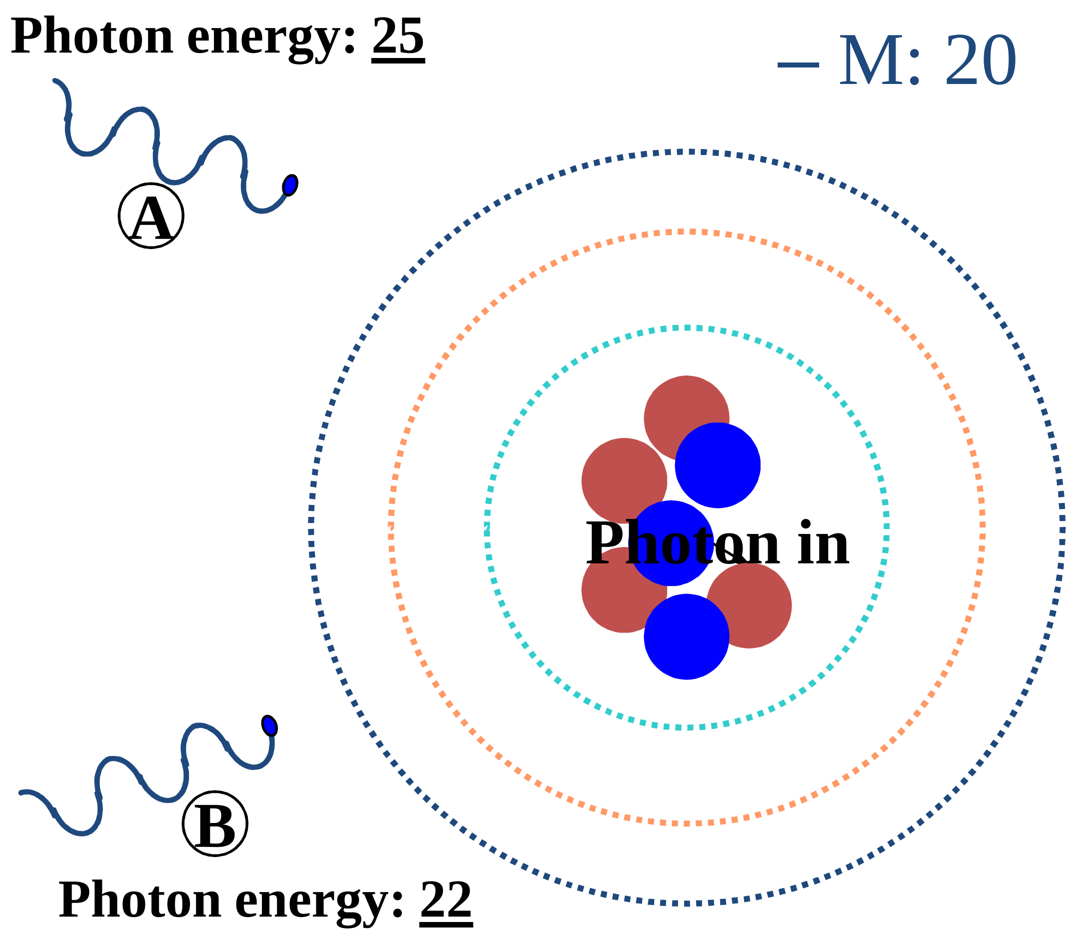
- inversely proportional to cube of photon energy
- proportional to cube of atomic number
- more likely with inner (higher) shells
- tightly bound electrons
- Interaction much more likely for
  - low energy photons
  - high atomic number elements

$$\text{P.E.} \sim \frac{1}{\text{energy}^3}$$

$$\text{P.E.} \sim Z^3$$

Photoelectric Threshold

- Binding Energies
  - K: 100
  - L: 50
  - M: 20



$$\text{P.E.} \sim \frac{1}{\text{energy}^3}$$

Which photon has a greater probability for photoelectric interactions with the m shell?

Photoelectric Threshold

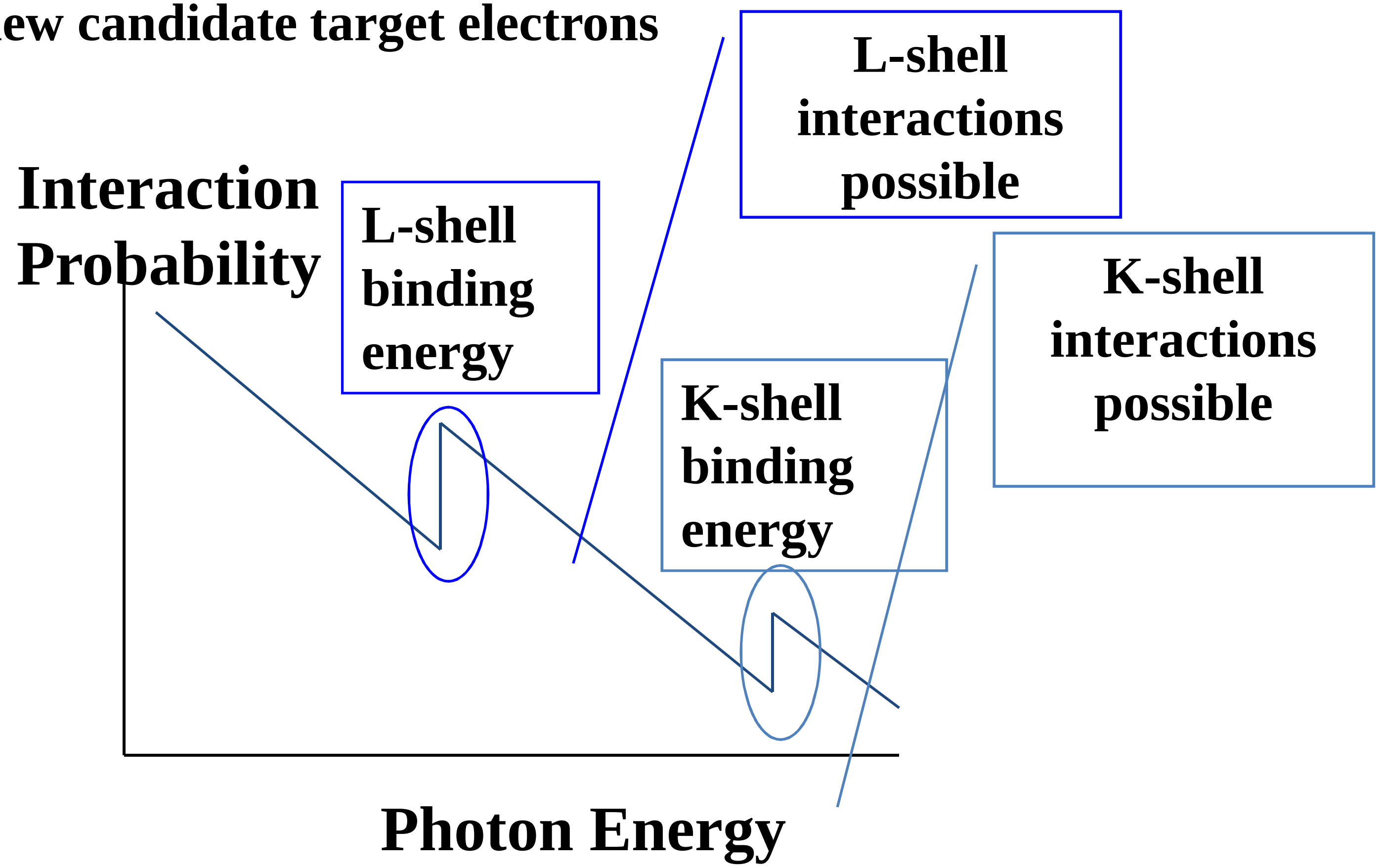
$$\text{P.E.} \sim \frac{1}{\text{energy}^3}$$

- **Photoelectric interactions decrease with increasing photon energy**

  BUT …

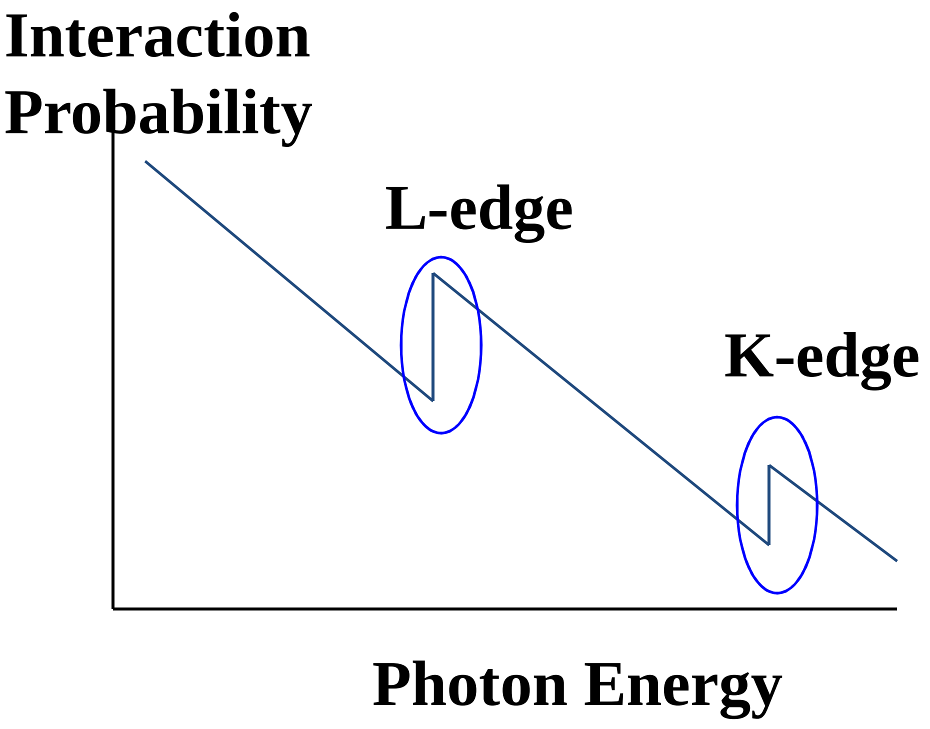
Photoelectric Threshold

- **When photon energies just reaches binding energy of next (inner) shell, photoelectric interaction now possible with that shell**
  - ♦ **shell offers new candidate target electrons**



Photoelectric Threshold

- **causes step increases in interaction probability as photon energy exceeds shell binding energies**



Photoelectric Effect

Why is this important?

- photoelectric interactions provide subject contrast
  - variation in x-ray absorption for various substances
- photoelectric effect does not contribute to scatter
- photoelectric interactions deposit most beam energy that ends up in tissue
  - **always** use highest kVp technique consistent with imaging contrast requirements

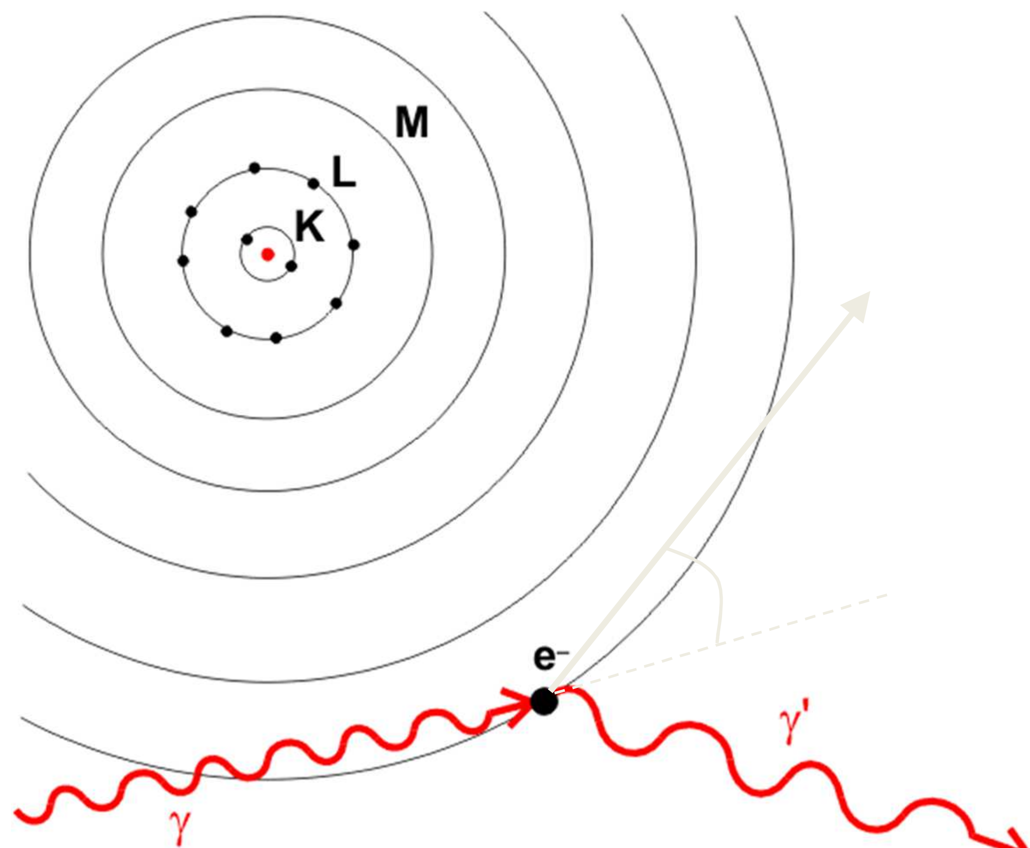
Compton Scattering

Compton scattering is dominant for intermediate photon energies.

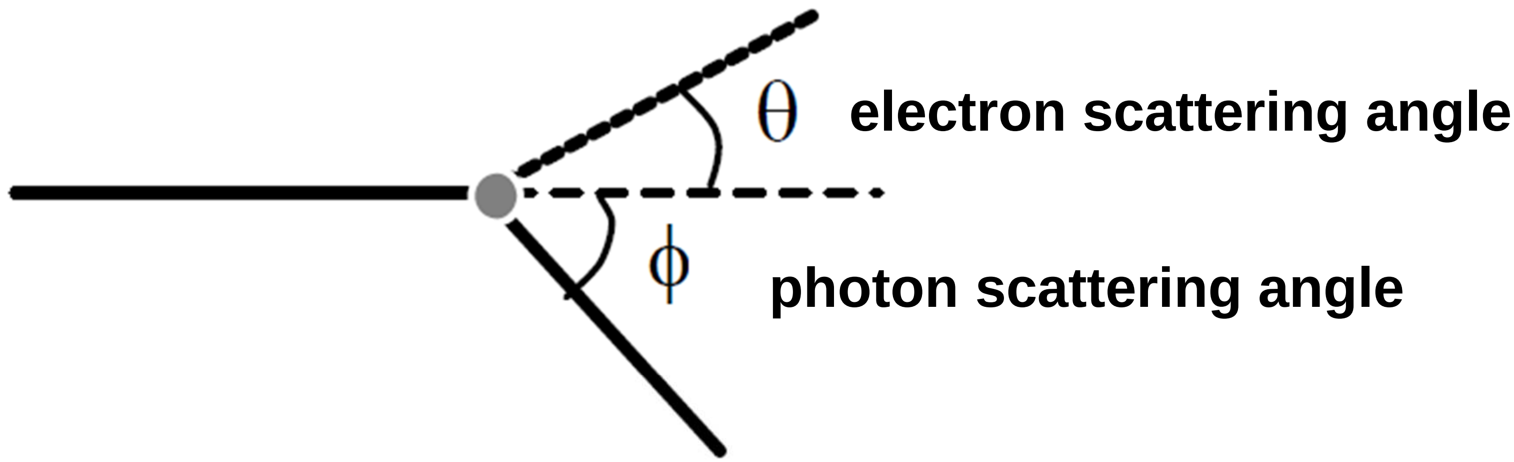
1. **Photon (γ) interacts with outer orbital electron.**

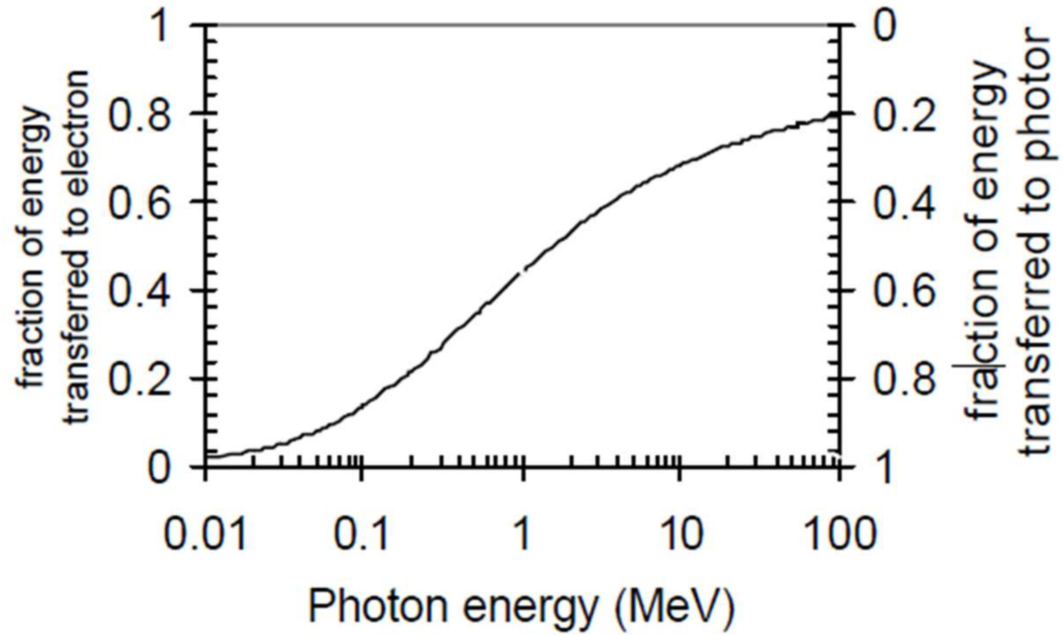
2. **Photon is scattered after transferring energy to the electron, which is ejected from the atom.**

3. **The scattered photon (γ') leaves at a different angle with less energy.**



Compton Scattering

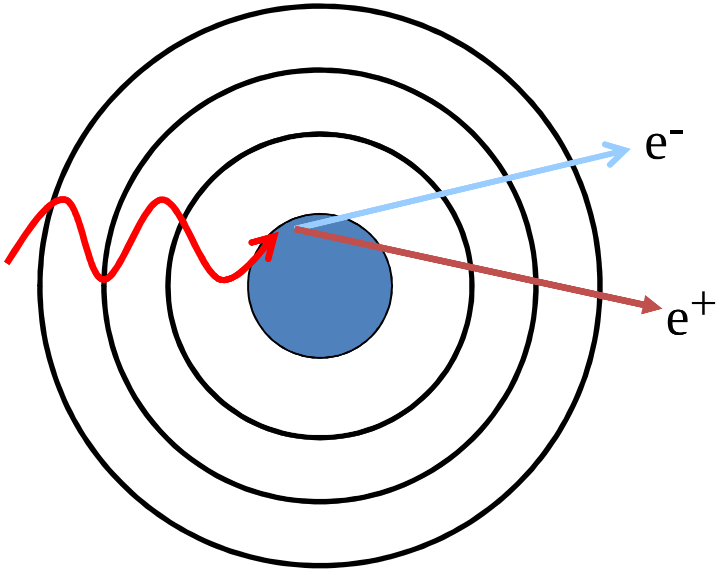


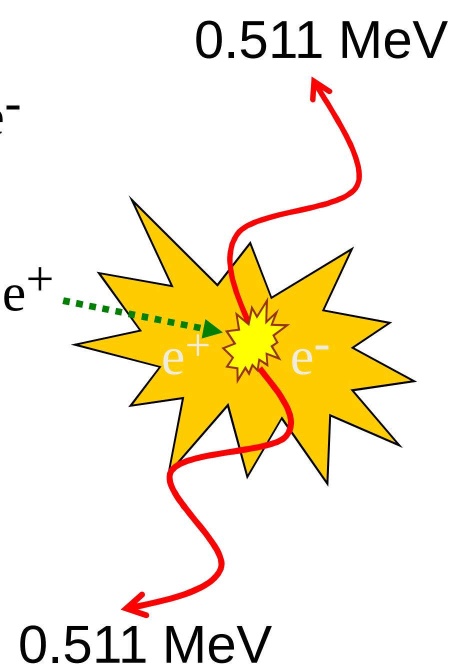


Pair Production

Must occur in the close vicinity of a nucleus. The incoming photon is absorbed and an electron-positron pair appears.

Requires minimum incoming photon energy of 1.022 MeV (0.511 MeV for the electron + 0.511 MeV for the positron)

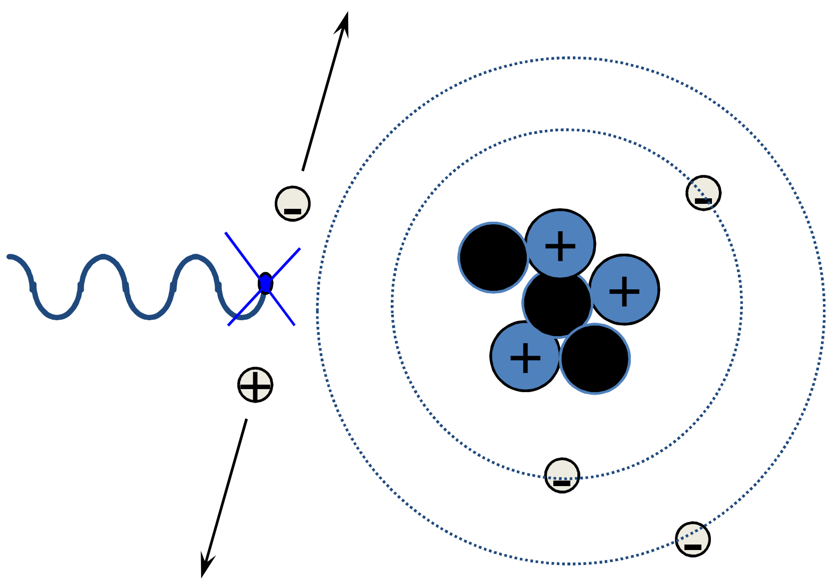




Positron ultimately combines with a stationary electron. They annihilate to produce two photons, each having 0.511 MeV energy and travelling in opposite directions

Pair Production

- Threshold energy for occurrence: 1.02 MeV
  - energy equivalent of rest mass of 2 electrons
- Threshold is above diagnostic energies
  - <u>does not occur</u> in diagnostic radiology



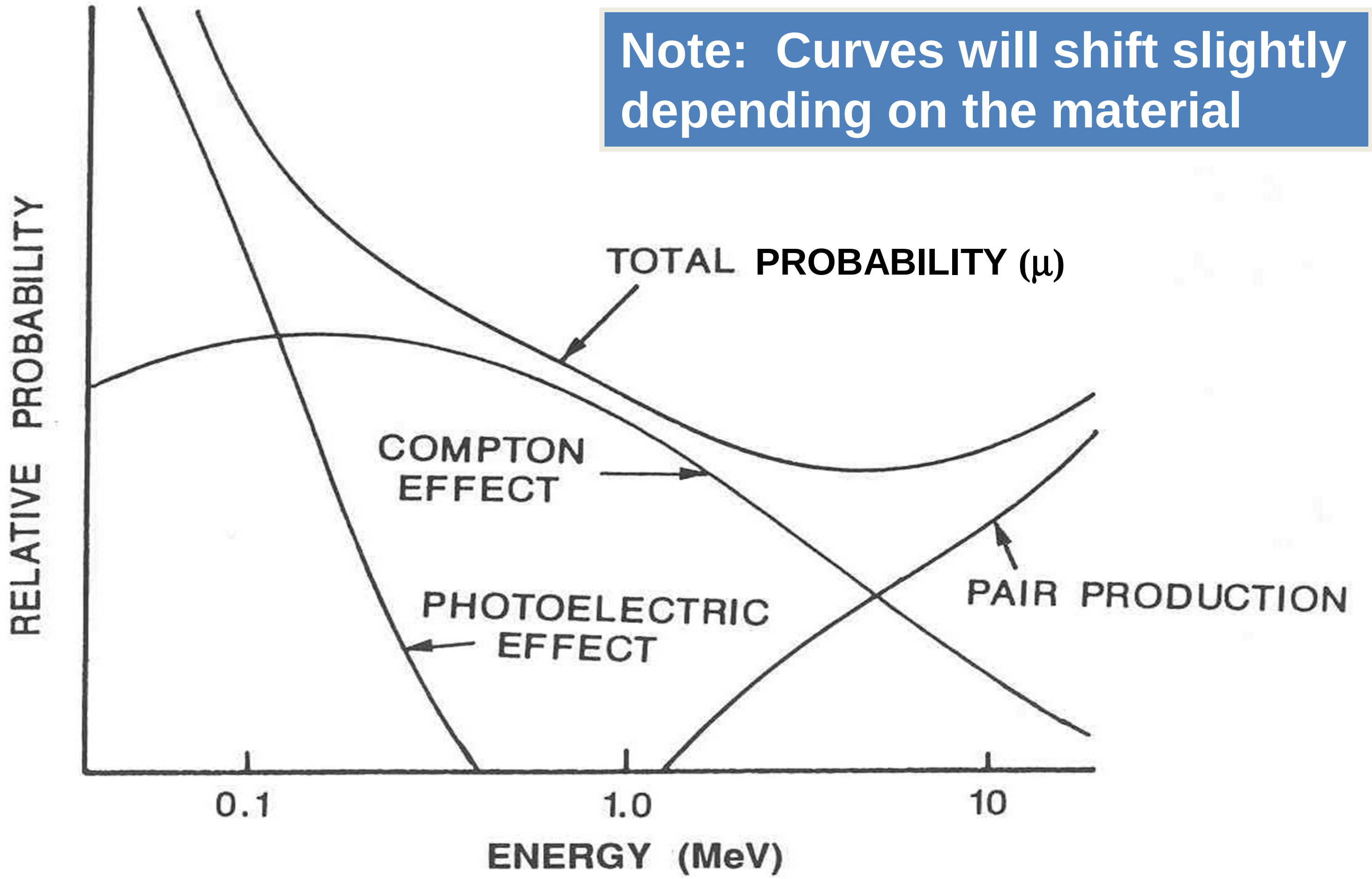
Linear Attenuation Coefficient

The linear attenuation coefficient (μ) is the total probability that a photon will interact as it travels through a material (units of cm^{-1}).

It is the sum of the probabilities of the different photon interactions occurring:

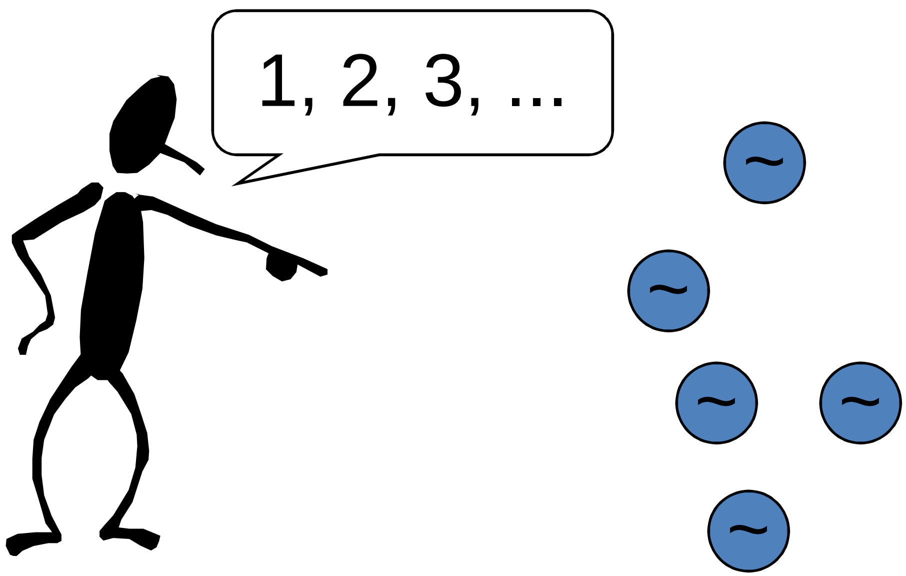
$$\mu = \mu_{PE} + \mu_{CS} + \mu_{PP}$$

Photon Interactions with Matter



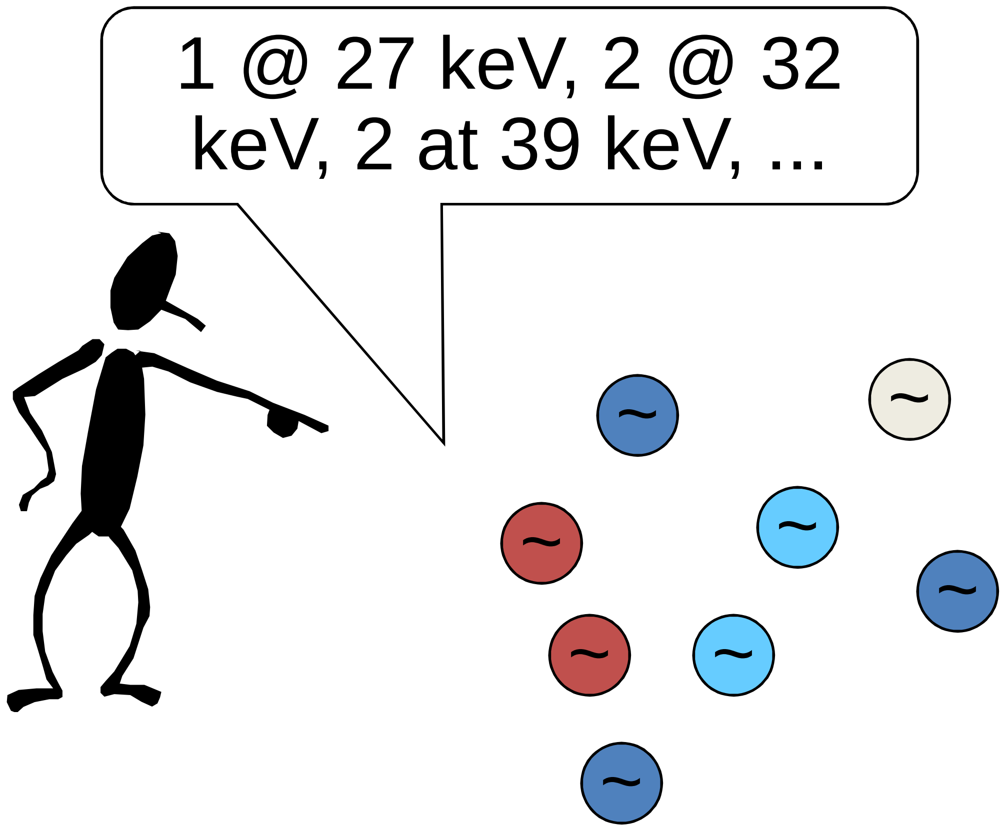
Beam Characteristics

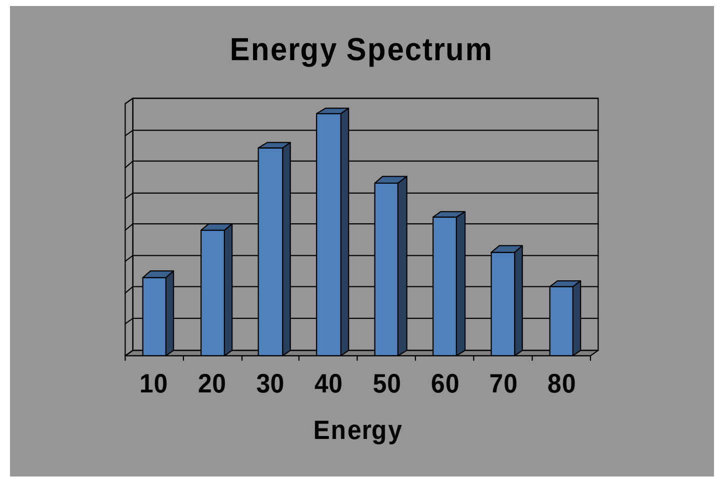
- **Quantity**
  - –<u>number</u> of photons in beam



Beam Characteristics

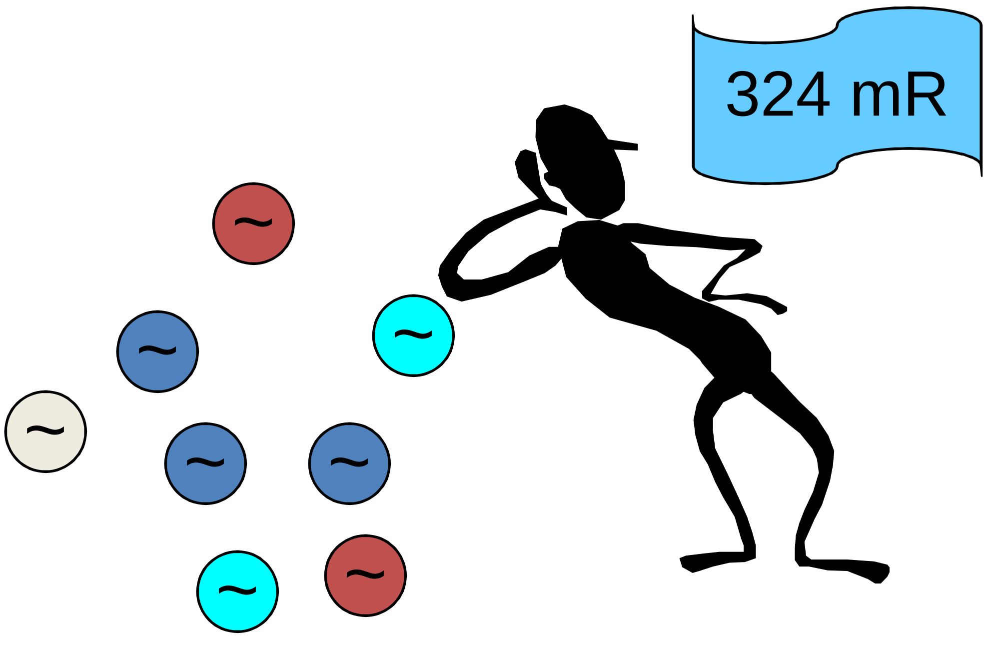
- **Quality**
  - –energy distribution of photons in beam





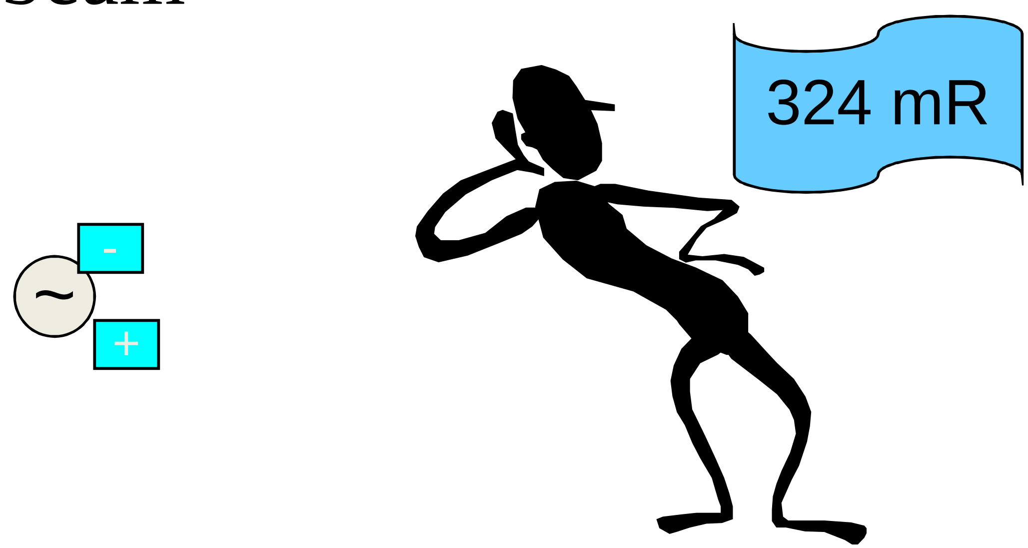
Beam Characteristics

- **Intensity**
  - weighted product of number and energy of photons
  - depends on
    - quantity
    - quality



Beam Intensity

- Can be measured in terms of # of ions created in air by beam
- Valid for monochromatic or for polychromatic beam



Attenuation Coefficient

- Parameter indicating fraction of radiation attenuated by a given absorber *thickness*
- Attenuation Coefficient is function of
  - absorber
  - photon energy

Monochromatic radiation beam

Linear Attenuation Coef.

- Why called linear?
  - distance expressed in linear dimension "x"
- Formula

$$N = N_o e^{-\mu x}$$

where

N_o = number of incident photons

N = number of transmitted photons

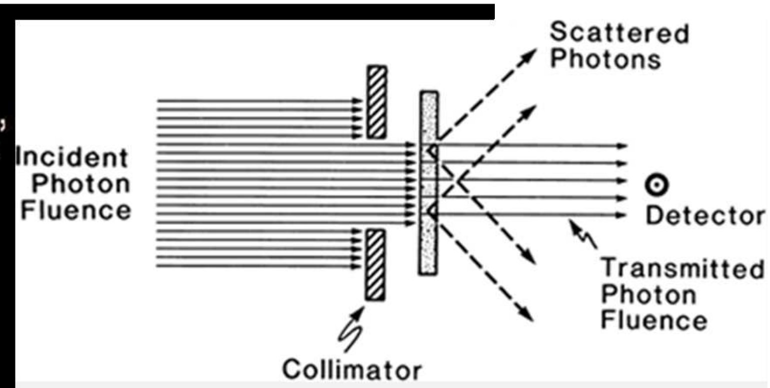
e = base of natural logarithm (2.718…)

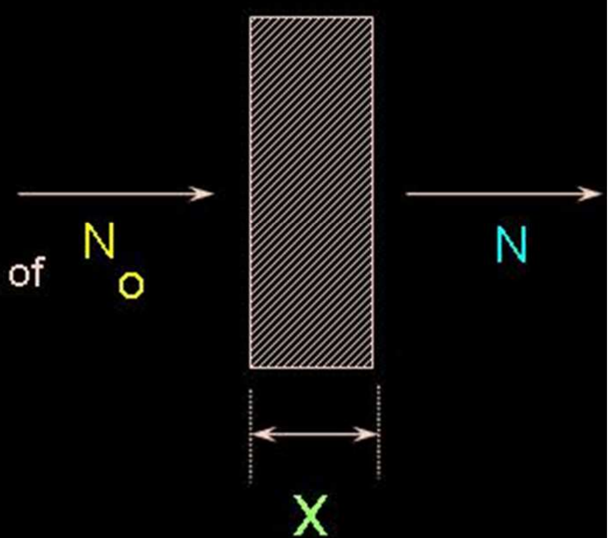
μ = linear attenuation coefficient (1/cm); property of

energy

material

x = absorber thickness (cm)





Monochromatic radiation beam

Linear Attenuation Coef.

Larger Coefficient = More Attenuation

$$N = N_o e^{-\mu x}$$

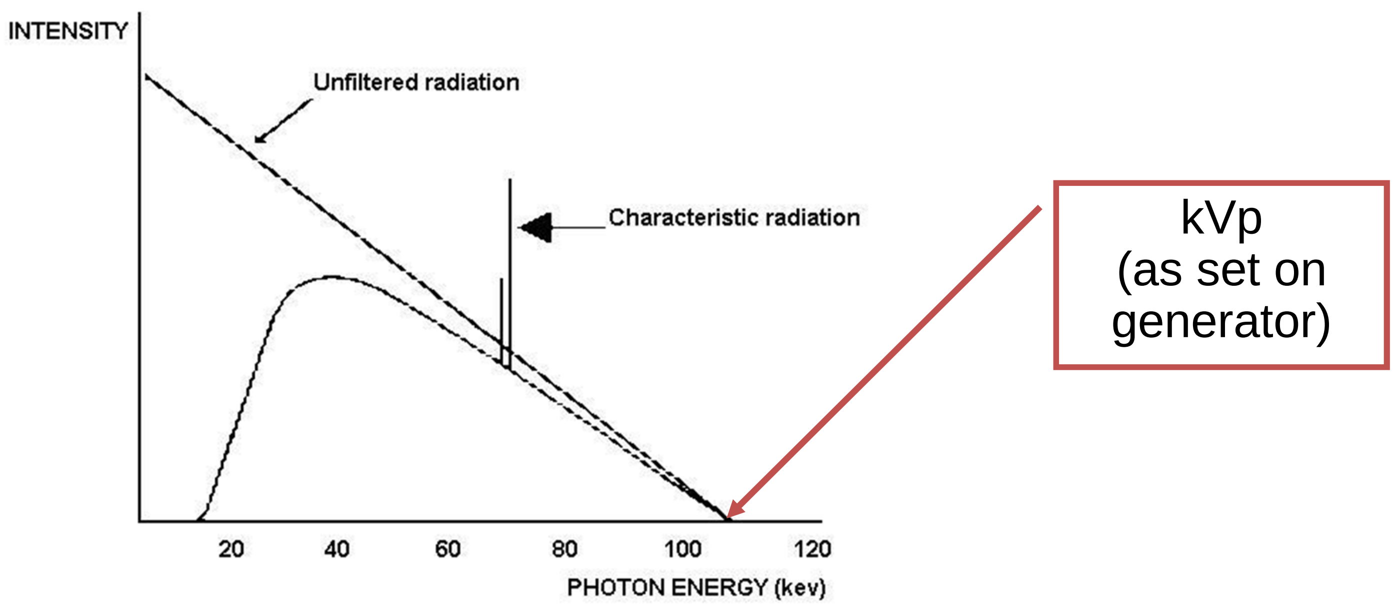
- Units:

  1 / cm (or 1 / distance)
- Properties
  - reciprocal of absorber thickness that reduces beam intensity by e (~2.718…)
    - ~63% reduction
    - 37% of original intensity remaining
  - as photon beam energy increases
    - penetration increases / attenuation decreases
    - attenuating distance increases
    - linear attenuation coefficient decreases
- Note: Same equation as used for radioactive decay

Monochromatic radiation beam

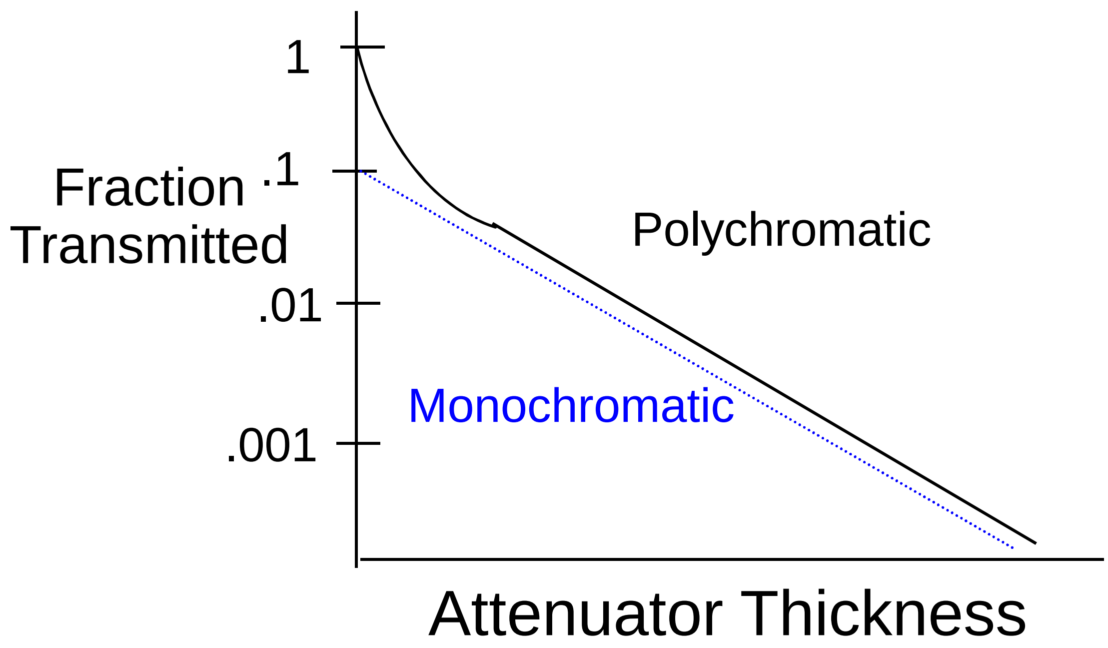
Polychromatic Radiation

- X-Ray beam contains spectrum of photon energies
  - –highest energy = peak kilovoltage applied to tube
  - –mean energy 1/3 - 1/2 of peak
    - depends on filtration



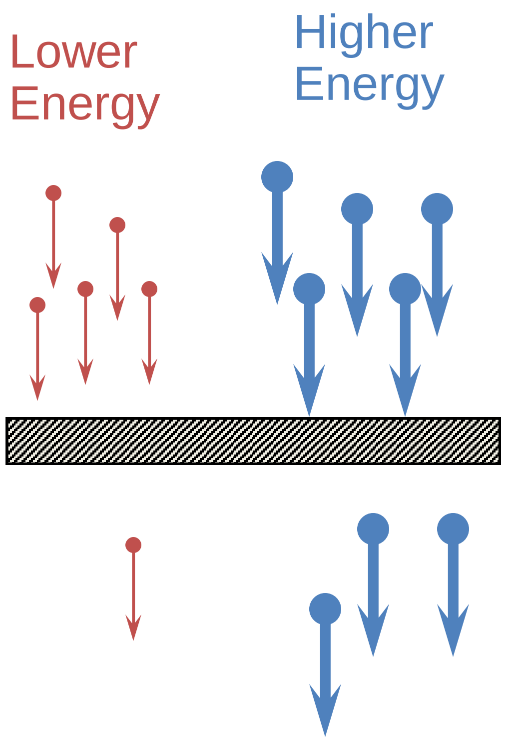
Polychromatic Attenuation

- Yields curved line on semi-log graph
  - line straightens with increasing attenuation
  - slope approaches that of monochromatic beam at the peak energy
- mean energy increases with attenuation
  - beam hardening



X-Ray Beam Attenuation

- reduction in beam intensity by
  - <u>absorption</u> (photoelectric)
  - <u>deflection</u> (scattering)
- Attenuation alters beam
  - quantity
  - quality
    - higher fraction of low energy photons removed
    - **Beam Hardening**



Half Value Layer (HVL)

$$N = N_o e^{-\mu x}$$

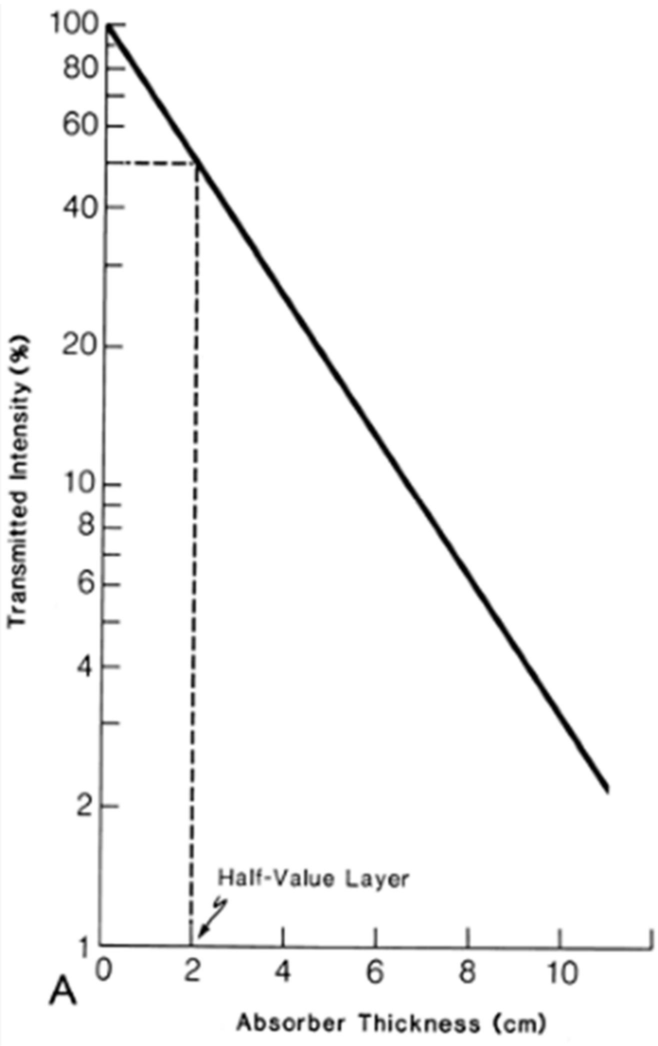
- absorber thickness that reduces beam intensity by exactly half
- Units of thickness
- value of "x" which makes N equal to $N_o / 2$

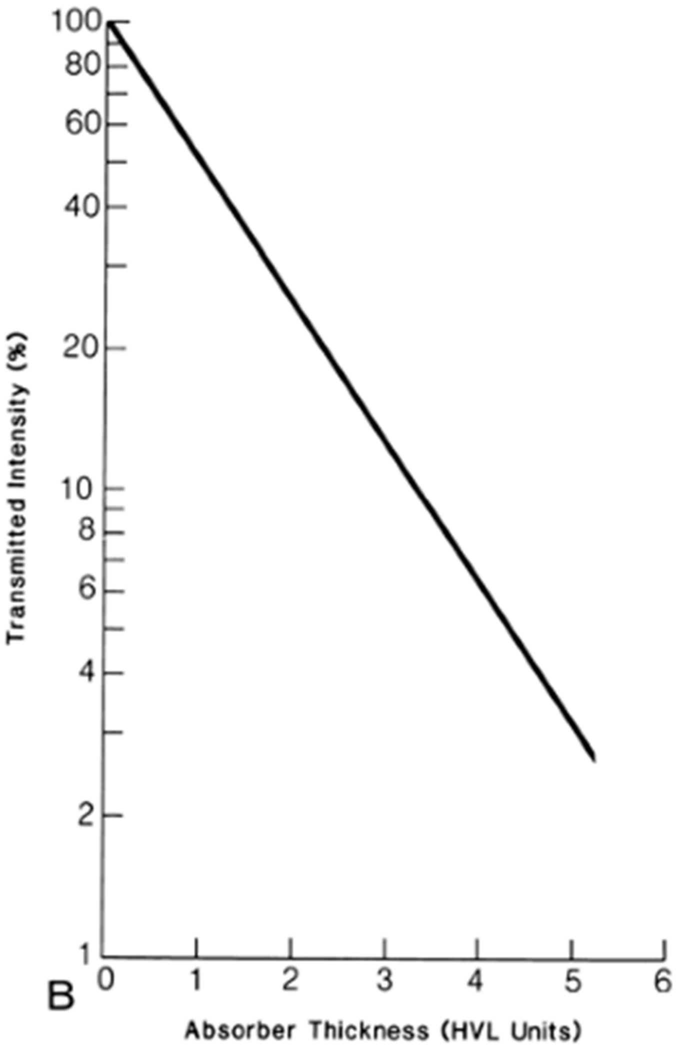
$$HVL = .693 / \mu$$

- Indication of beam quality
- Valid concept for all beam types
  - Mono-energetic
  - Poly-energetic
- Higher HVL means
  - more penetrating beam
  - lower attenuation coefficient

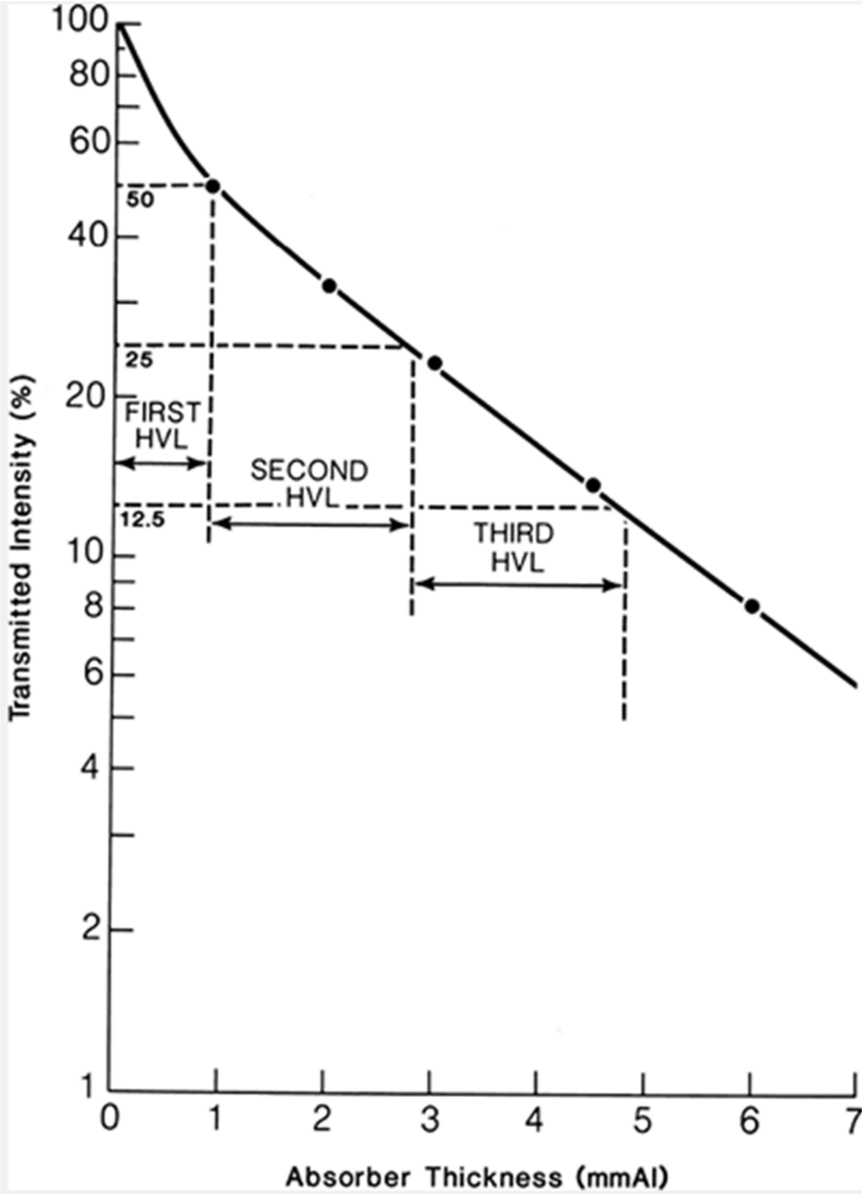
Half Value Layer (HVL)

HVL = 2 cm and Âµ = 0.347 cm-1





Half Value Layer (HVL)



Spectrum of photon energies
-aluminum absorber
First HVL = 0.99 mm
Second HVL = 1.9 mm
Third HVL = 2.0 mm

The first HVL < subsequent HVLs.

Filter thickness ↑

av. energy of the transmitted beam ↑

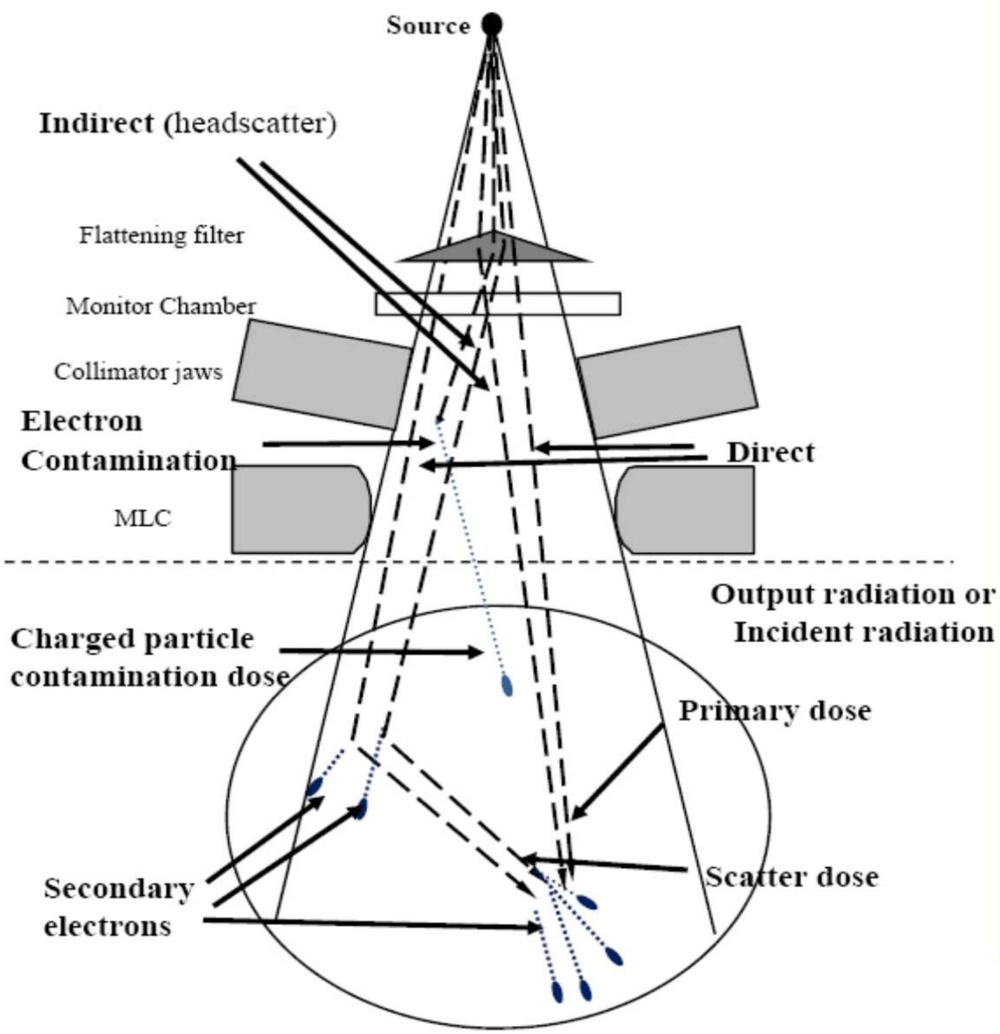
OR

the beam becomes ↑ harder

Factors Affecting Attenuation

- Energy of radiation / beam quality
  - higher energy
    - more penetration
    - **less** attenuation
- Matter
  - density
  - atomic number
  - electrons per gram
  - higher density, atomic number, or electrons per gram **increases** attenuation

Sources of radiation that determine dosimetric characteristics of clinical photon beams



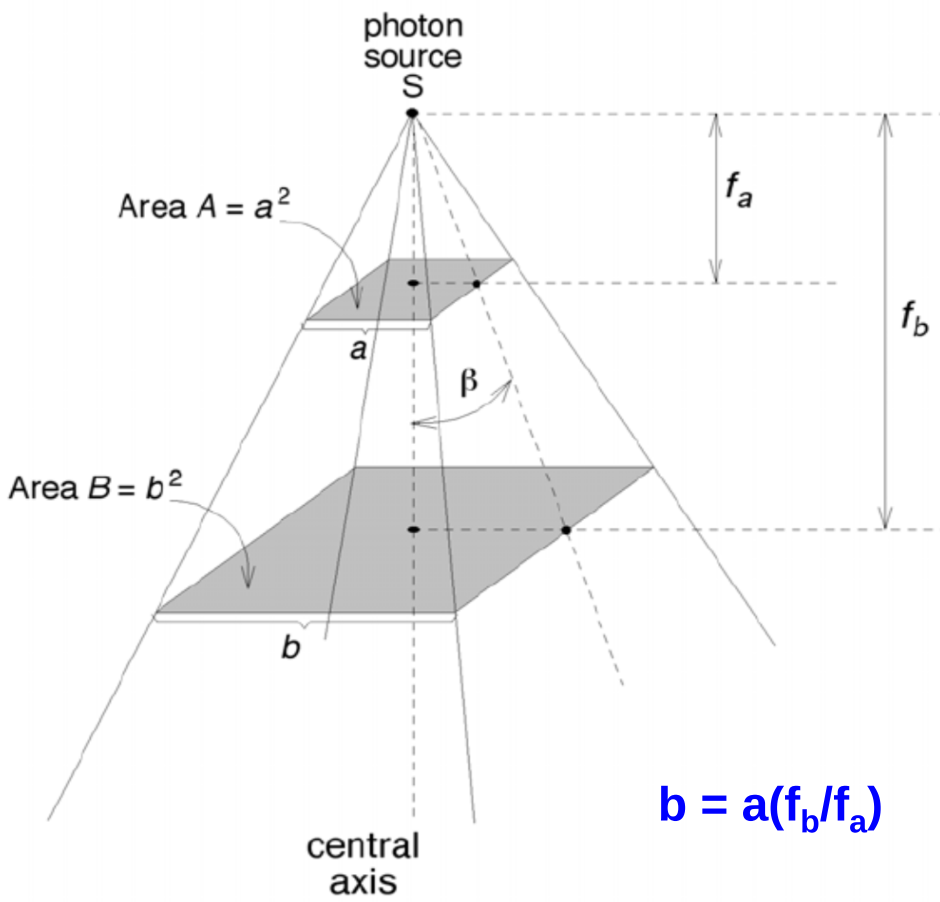
- **Direct Radiation (Focal Radiation)**
  - Photon radiation generated at the target that reaches patient without any intermediate interactions.
- **Indirect Radiation (Extra-focal Radiation):**
  - Photon radiation with a history of interaction/scattering in the head of the treatment unit with the flattening filter, collimators, or other structures in the treatment head .
- **Contaminant electrons/positrons**
  - secondary electrons and positrons released from interactions with either the treatment head or the air column .

AAPM TG74 Report

Inverse Square Law & Field Divergence

- Photon beam sources assumed to be point sources

- Beams produced are divergent



$b = a(f_b/f_a)$

Passage Through a Medium

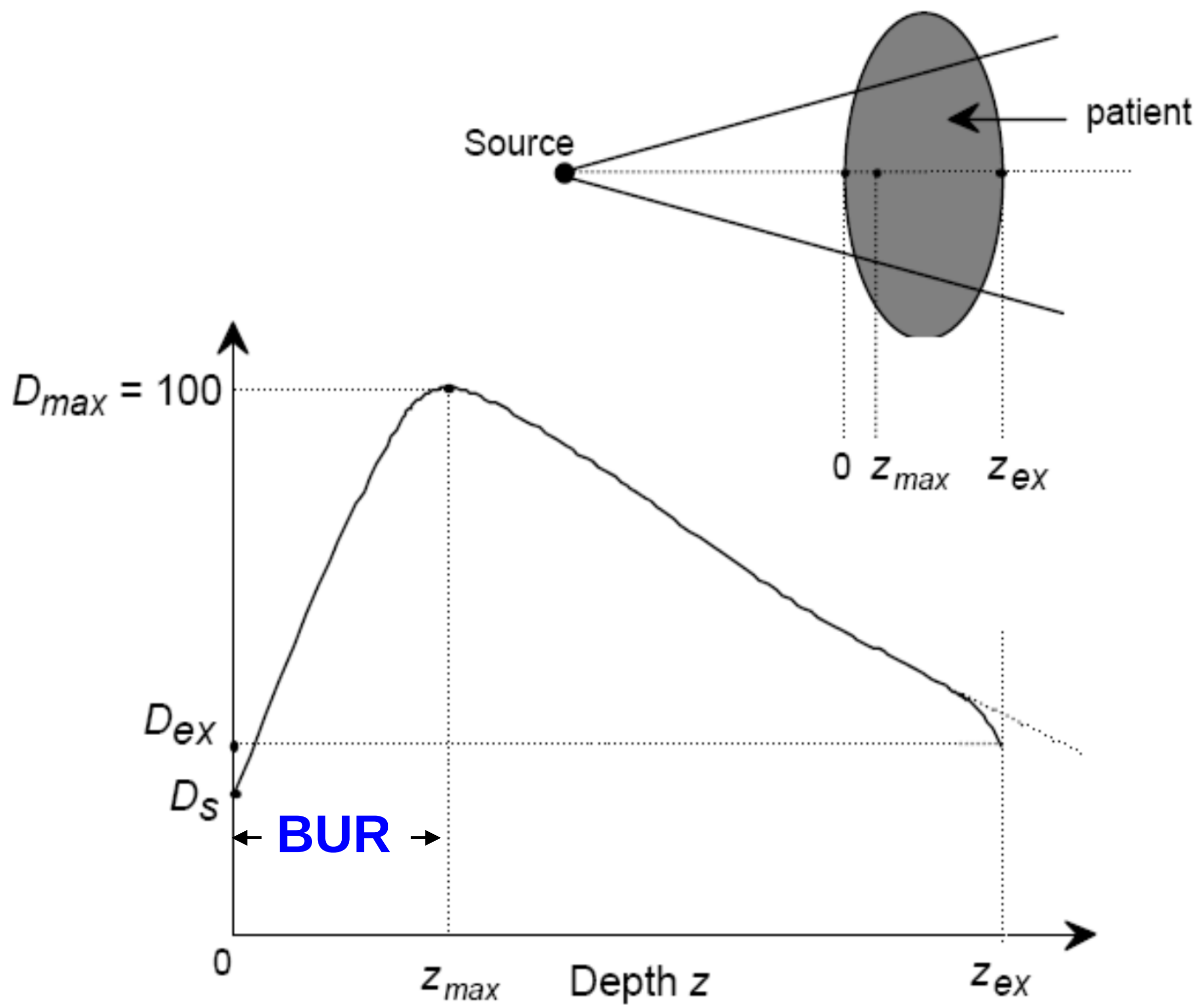
Z_{max} = depth of dose maximum (d_m)

D_{max} = Dose maximum

Z_{ex} = depth at exit surface (d_{ex})

D_{ex} = Exit dose

D_s = Surface dose



Dose buildup

Buildup of dose increases with increase in energy of the beam. The region between the surface and the point of maximum dose is called the dose buildup-region.

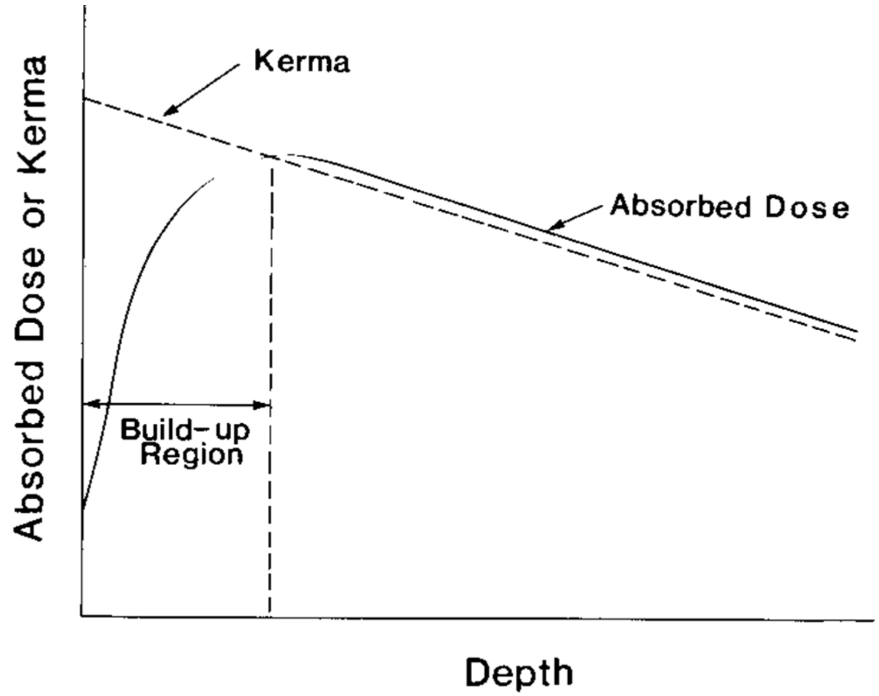


Figure 9.4. Schematic plot of absorbed dose and kerma as functions of depth.

- Kerma-- (1) kinetic energy released in the medium; (2) the energy transferred from photons to directly ionizing electron; (3) maximum at the surface and decreases with depth due to decreased in the photon energy fluence; (4) the production of electrons also decreases with depth

- Absorbed dose: (1) depends on the electron fluence;

  (2) high-speed electrons are ejected from the surface and subsequent layers;(3) theses electrons deposit their energy a significant distance away from their site of origin

Depth of dose maximum (d_m) and D_{ex}

d_m depends on

Beam energy, and

Field size

- dependence on beam energy

D_{ex}
Dose at exit surface
Depends on beam energy

Beam	d_m (cm)
Co-60	0.5
4 MV	1.0
6 MV	1.5
10 MV	2.5
15 MV	3.0
18 MV	3.5

Percentage depth dose (PDD)

Percentage depth dose is defined as the quotient, expressed as a percentage, of the absorbed dose at any depth d to the abosrbed dose at a fixed reference depth d_0 along the central axis of the beam

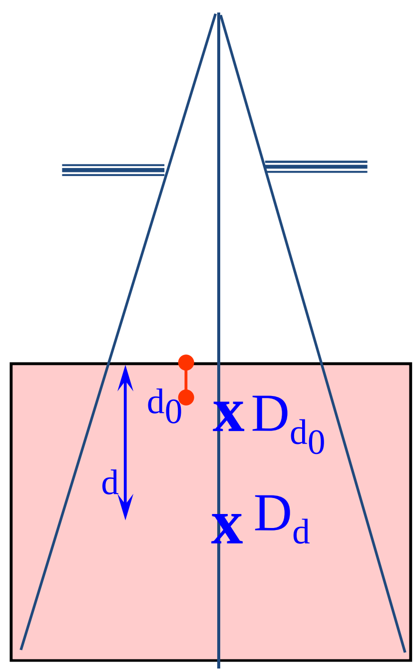
$$P = \frac{D_d}{D_{d_0}} \times 100$$

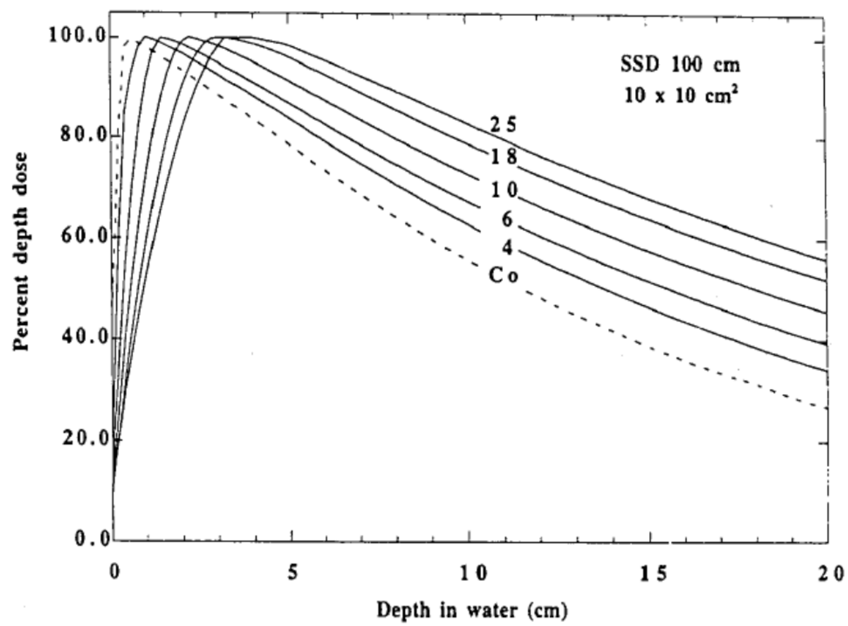
Orthovoltage (400kVp) d_0 = Surface
Higher energies $d_0 = d_m$

$$Dmax = \frac{D_d}{P} x100$$

PDD depends on

- beam energy
- depth
- field size
- distance from source
- beam collimator system



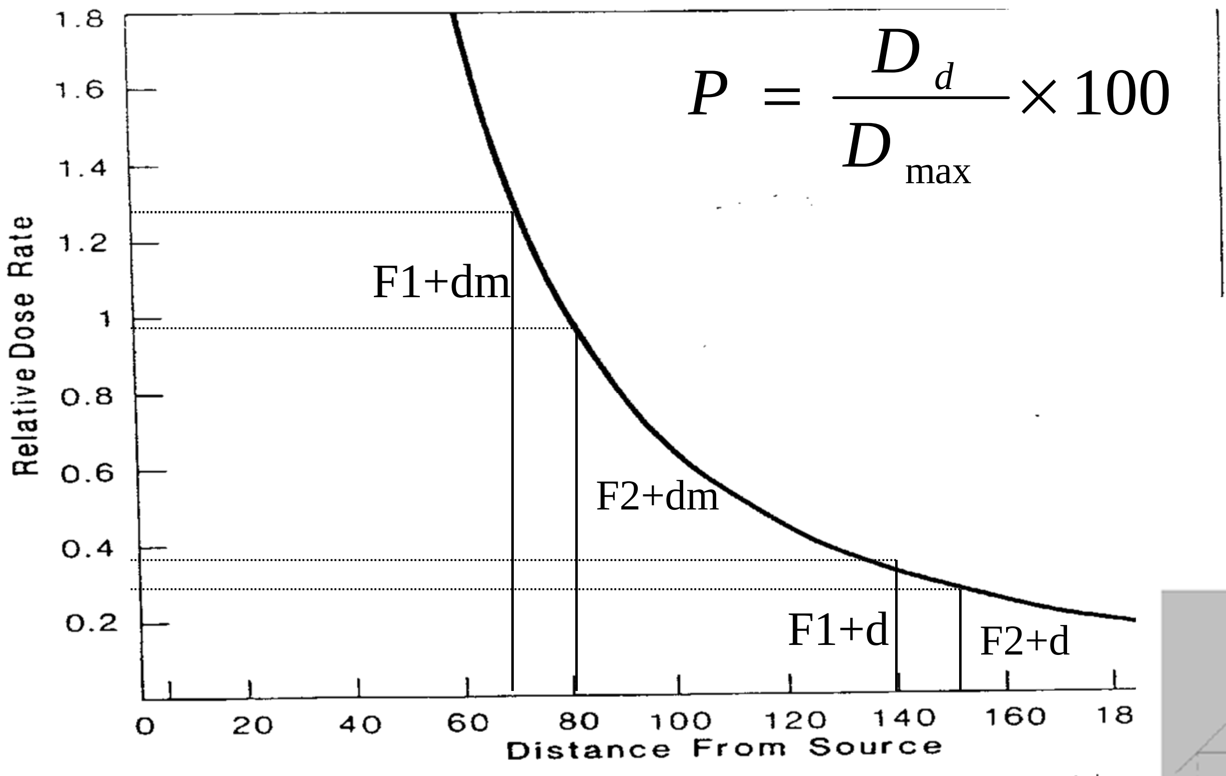


PDD dependence on SSD

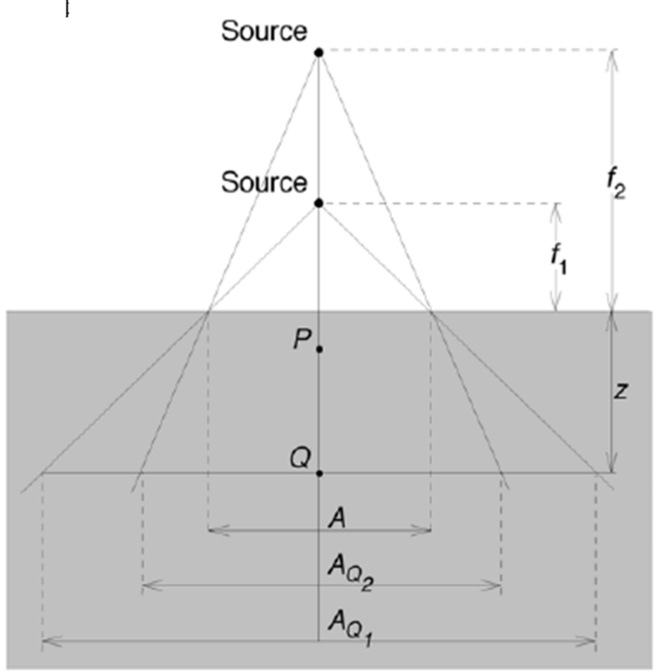
Mayneord F Factor

$$F = \left(\frac{f_2 + d_m}{f_1 + d_m}\right)^2 \times \left(\frac{f_1 + d}{f_2 + d}\right)^2$$

- Photon fluence from a point source varies inversely as a square of the distance from the source. (SSD> 80cm)
- PDD increase with SSD

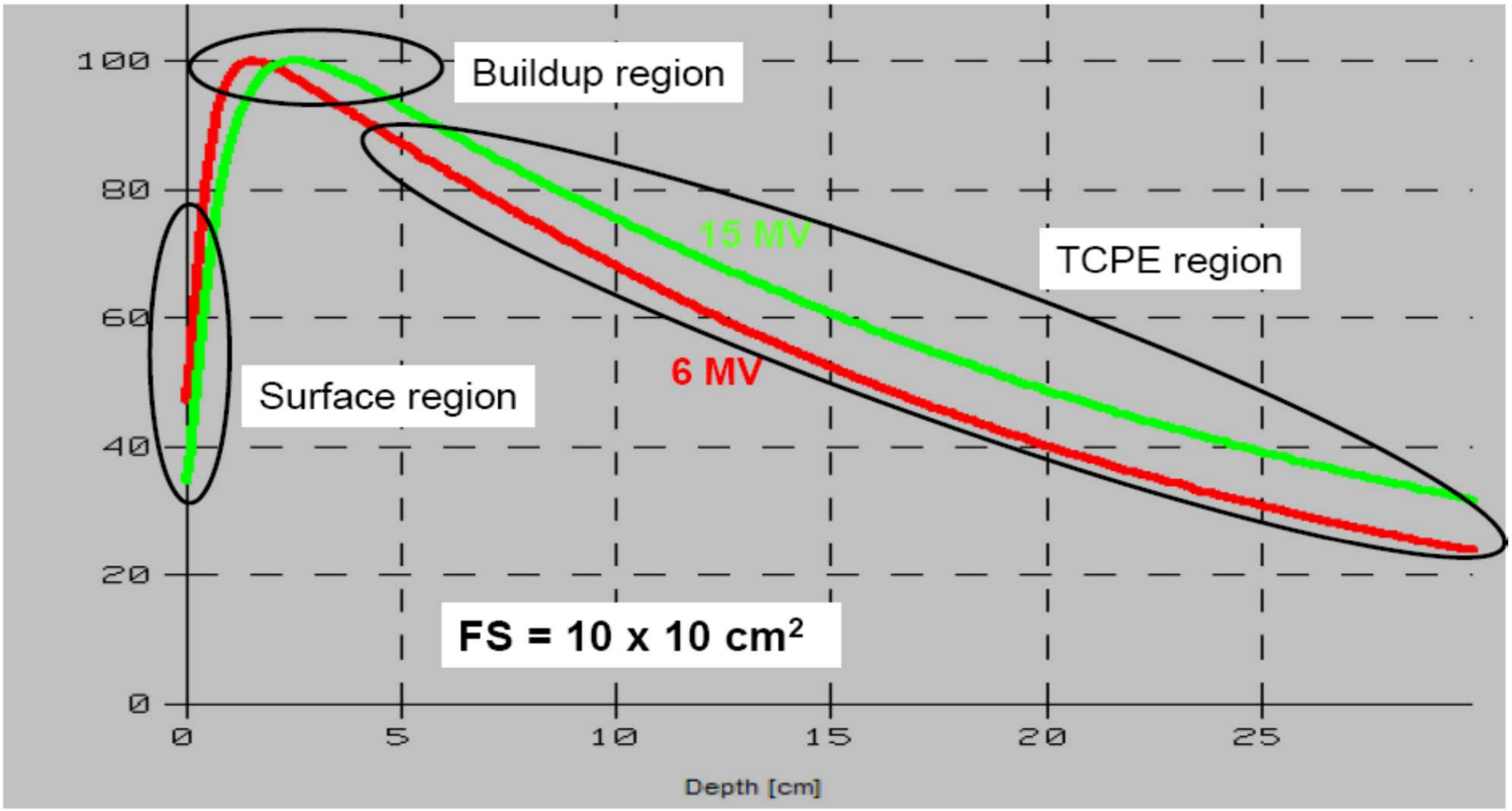


$$P = \frac{D_d}{D_{max}} \times 100$$



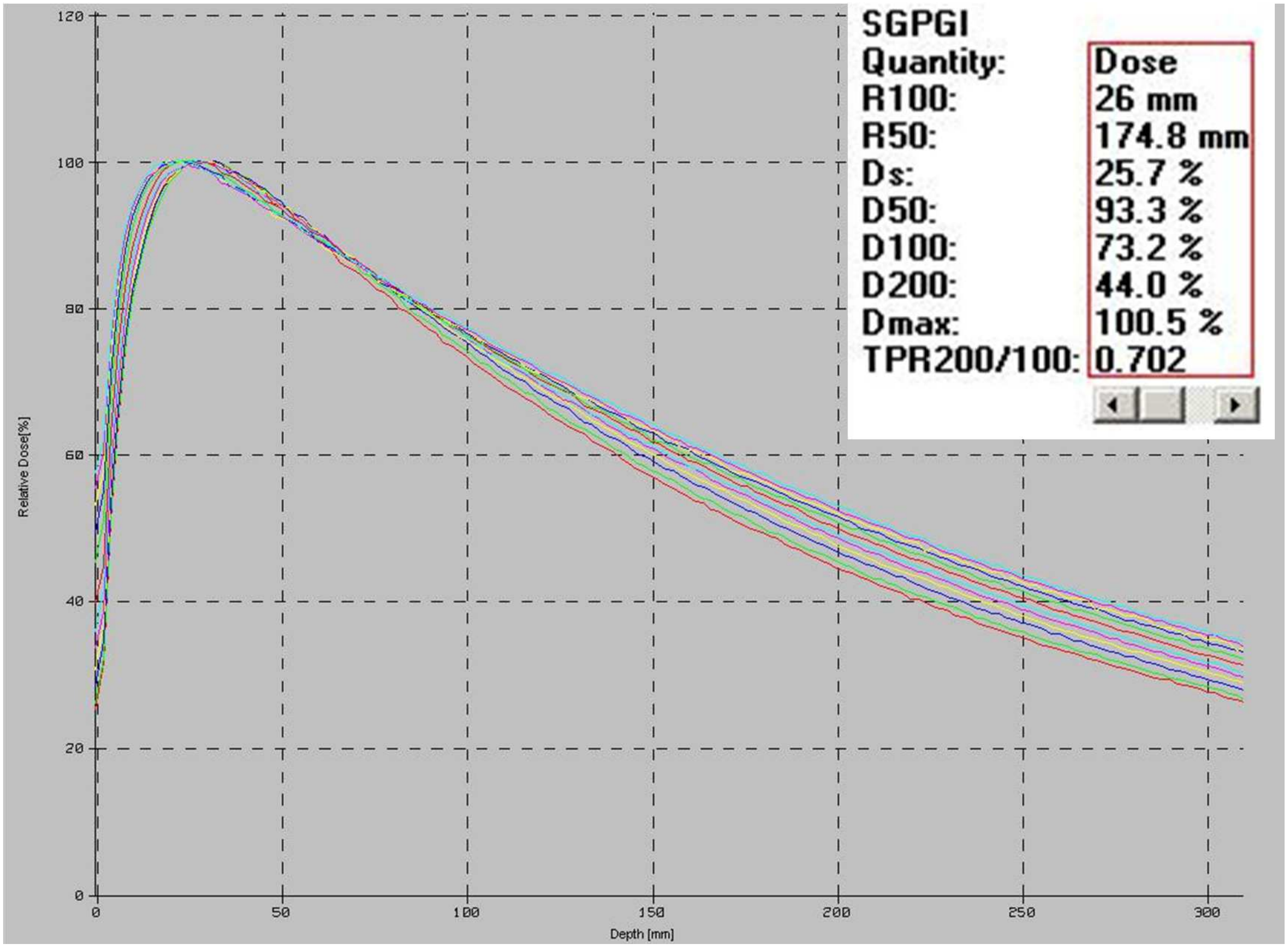
Normalized Depth Dose Data

Energy Dependence



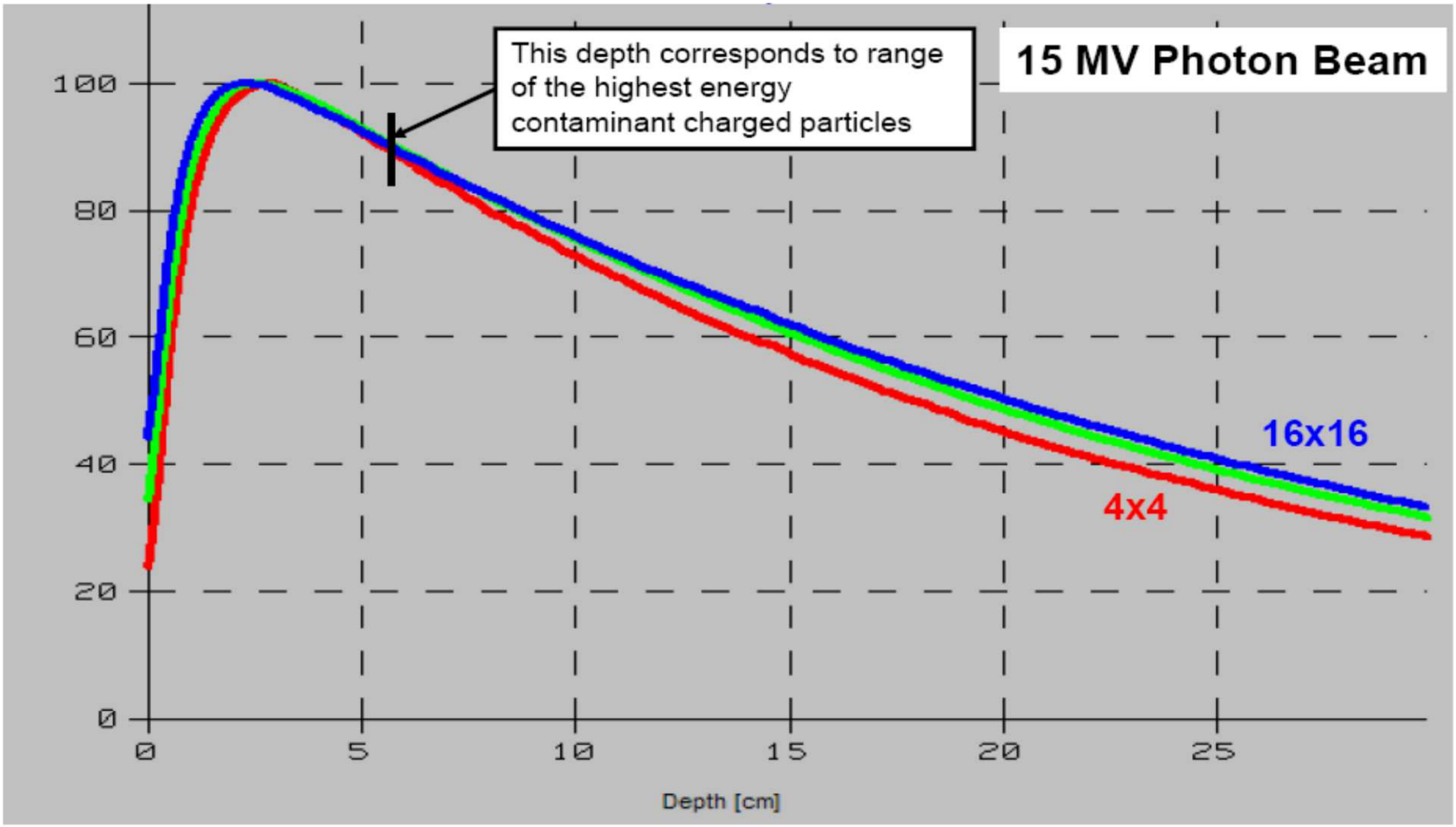
Percentage Depth Dose

Characteristics



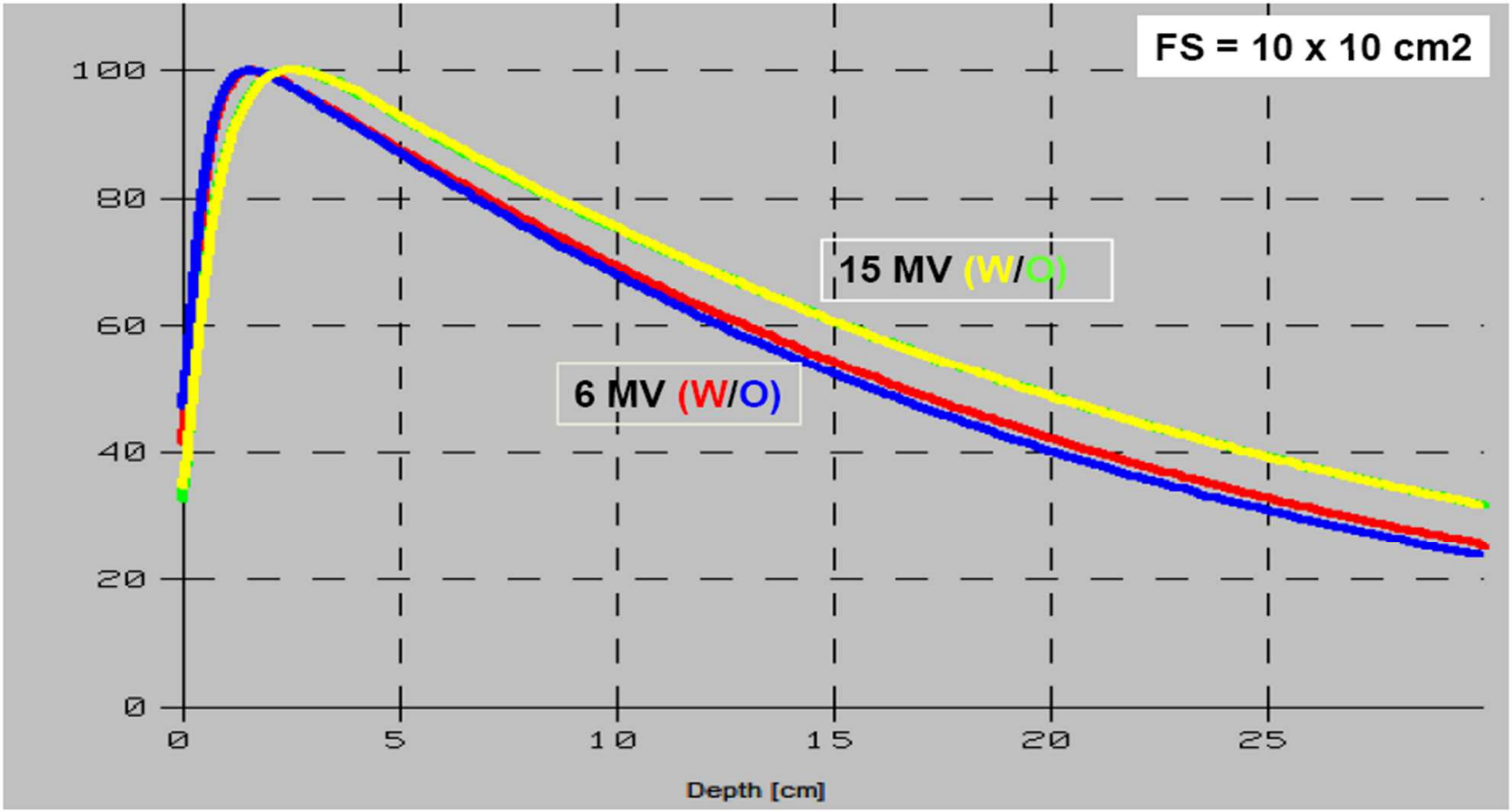
Percentage Depth Dose

Field Size Dependence



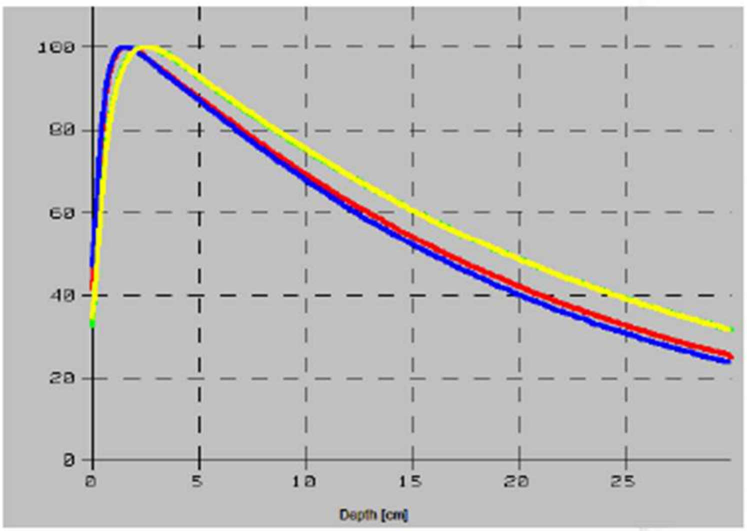
Percentage Depth Dose

Wedge/Open Comparison



Normalized Depth Dose Data

Wedge/Open Comparison



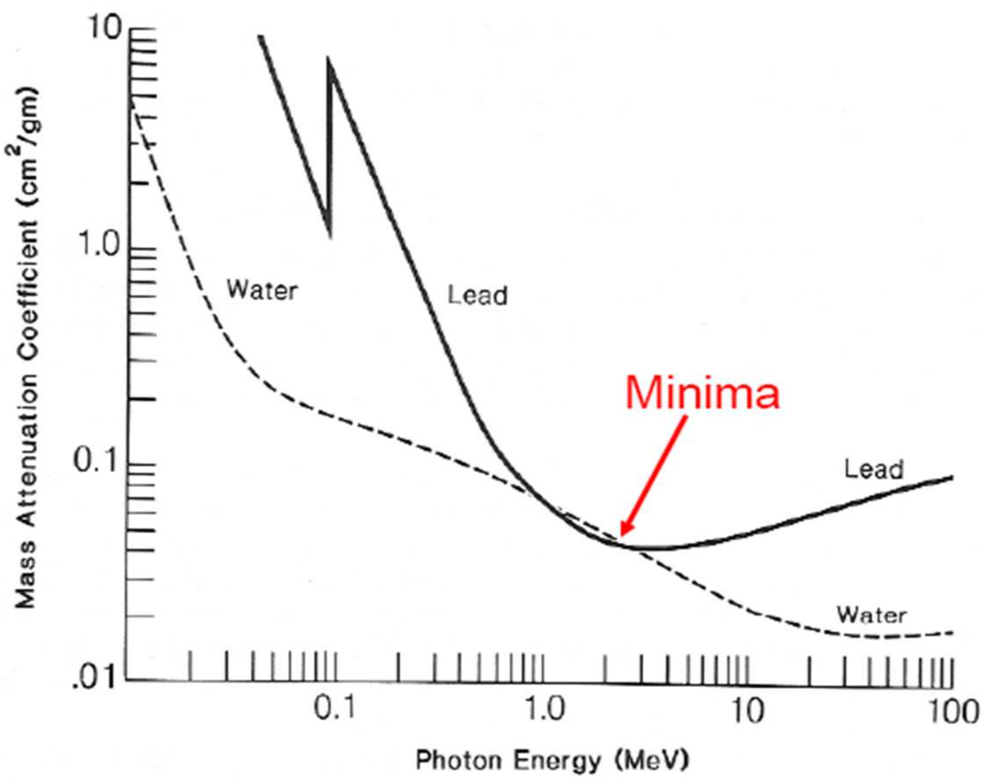
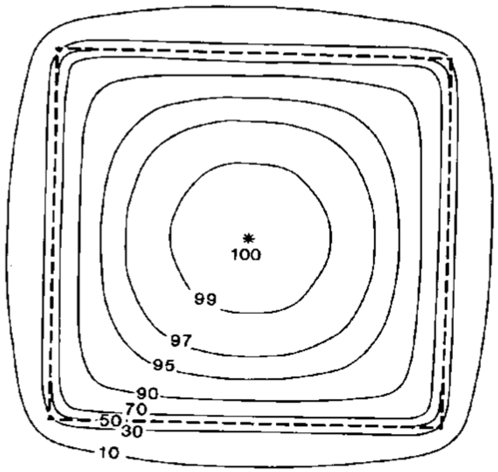


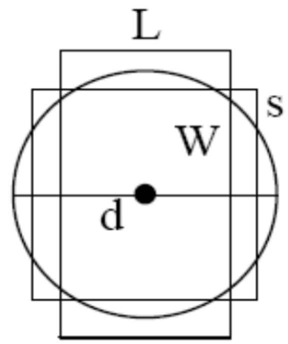
Figure 5.12. Plot of total mass attenuation coefficient (μ/ρ) as a function of photon energy for lead and water. Reprinted with permission from Johns HE, Cunningham JR. The physics of radiology. 3rd ed. Springfield, IL: Charles C Thomas, 1969.

Effect of field size and shape



- Geometrical field size
- Dosimetric (Physical) field size
- Field size increases the scatter increases. Scattered dose is greater at larger depth than at the depth of Dmax. PDD increases with increasing field size.
- The increase in PDD by increase in field size depends on beam quality.
- Field size dependence of PDD is less for higher energy than for lower energy beams
- PDD for rectangular field is calculated by area by perimeter approximation

Equivalent square



Sterling Formula:

(Sterling et.al., Brit. J. Radiol. 37, 544 (1964))

$$S = \frac{2LW}{L+W} = 4A/P$$

Assuming, λ = 0.26 cm-1., and μ = 0.5

$$S(L,W) = 4\int_0^{L/2}\int_0^{W/2} D(x,y)dxdy$$

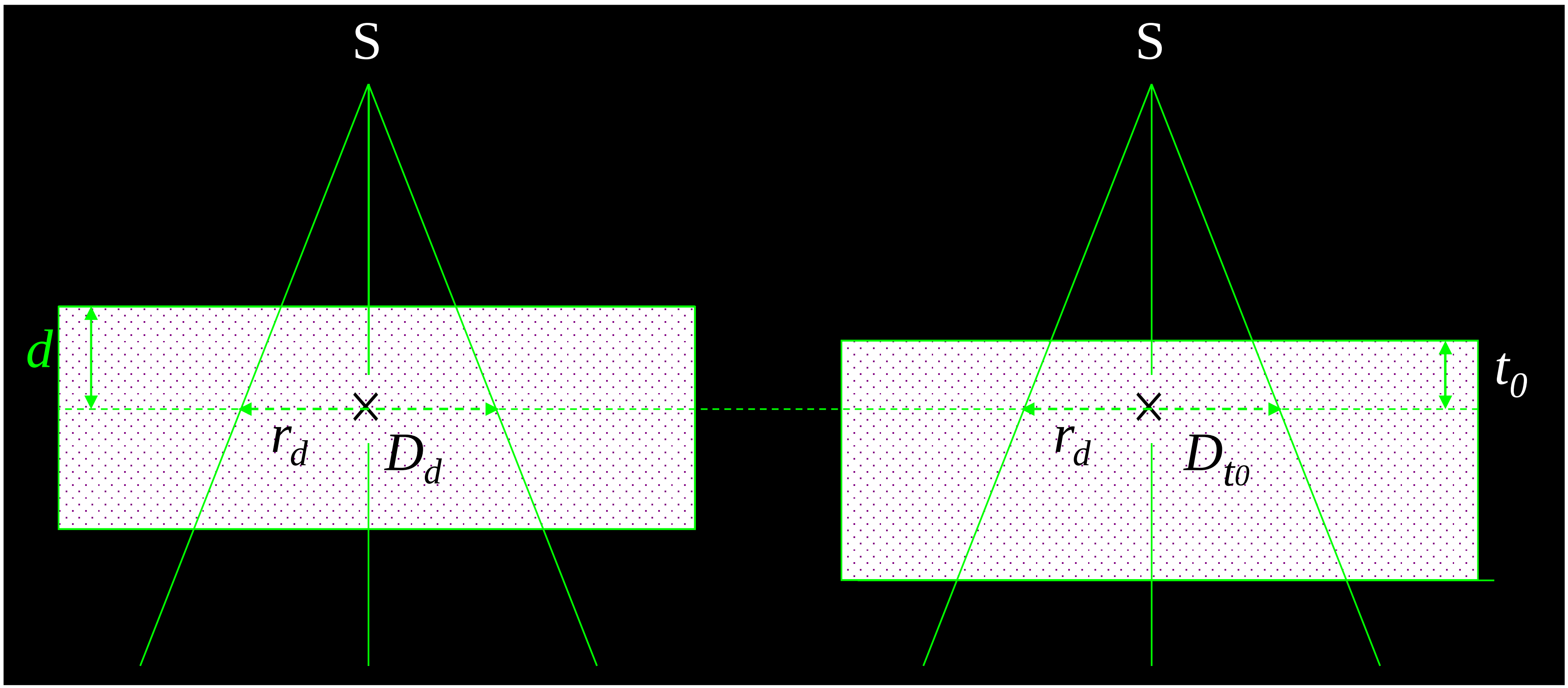
L/W	1	2	3	4	5
$S(L,W)/S(10,10)$	1.000	0.993	0.982	0.969	0.958

TMR and TPR

$$TMR(d, r_d) = \frac{D_d}{D_{t0}}$$

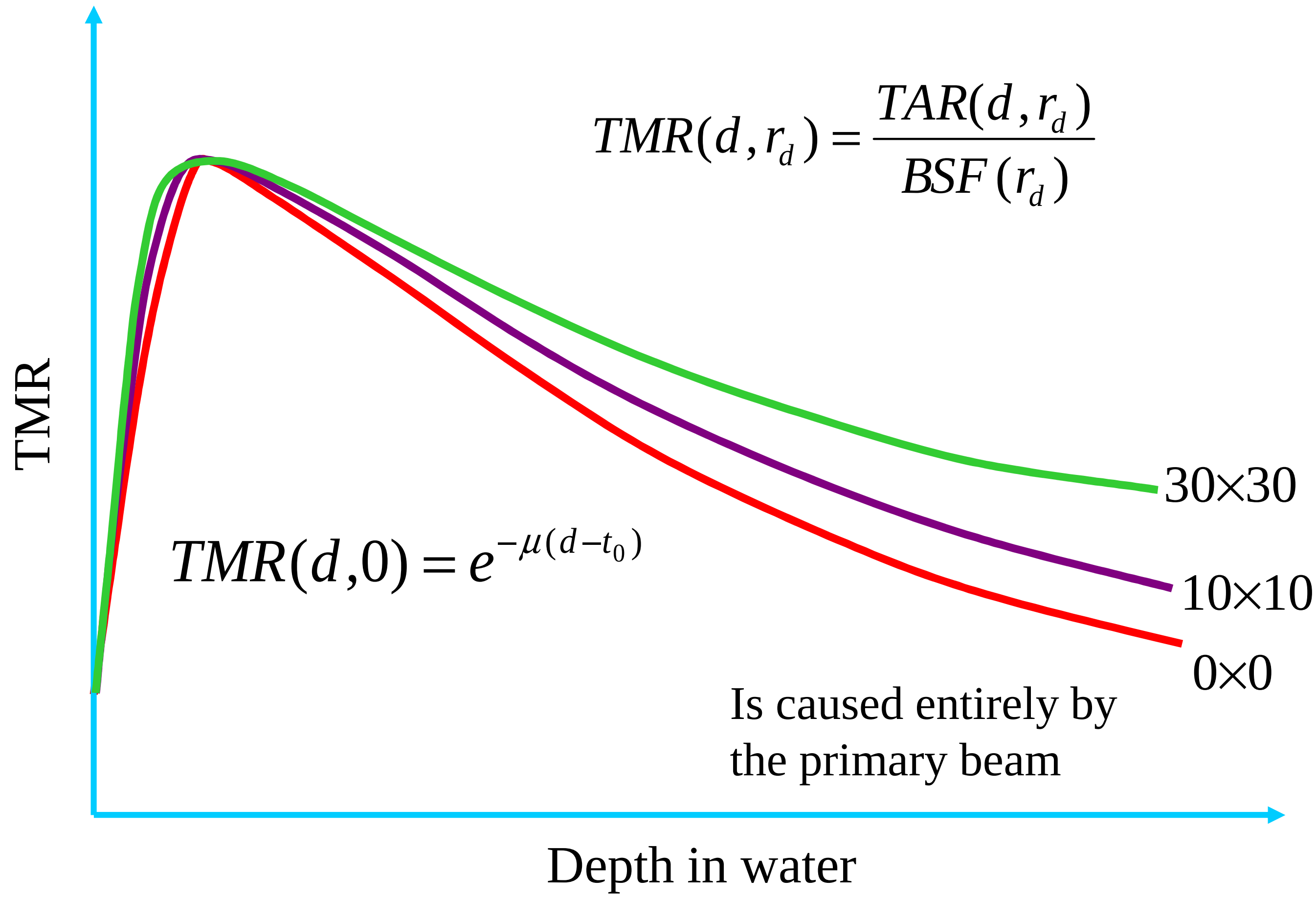
For TPR, $t_0 = d_{10}$

For TMR, $t_0 = dm$



Properties of TMR

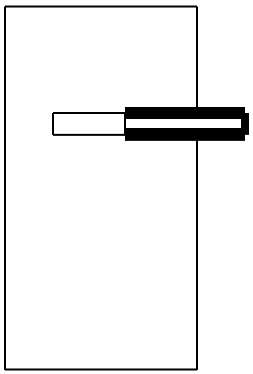
TMR is independent of SSD, increases with energy and field size.



TMR data for 10 MV x-ray beams

Collimator Scatter Factor (S_c)

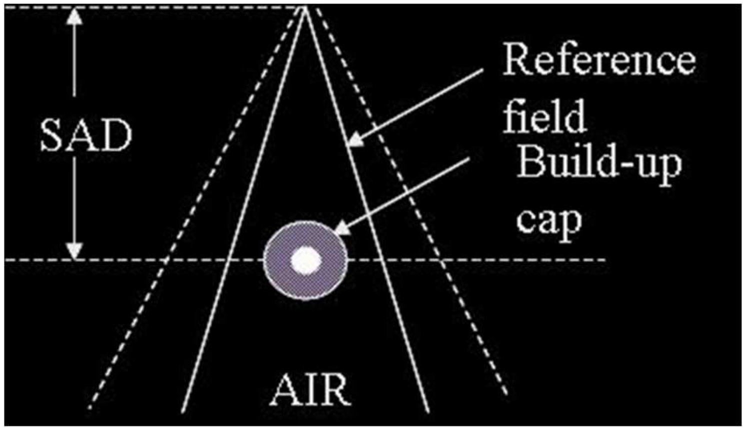
- The beam output measured in air depends on the field size
  - Field size ↑; output ↑; collimator scatter ↑
  - "Output factor"

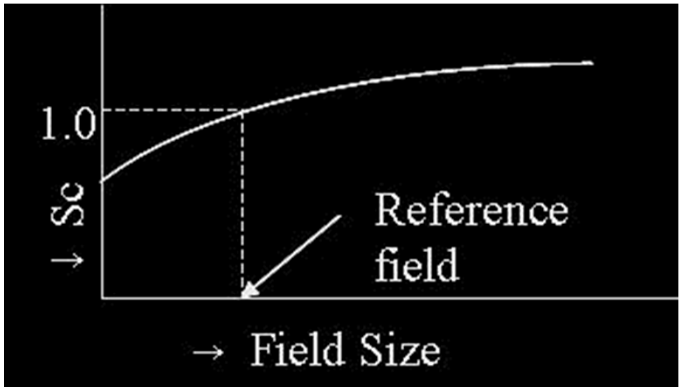


Mini phantom

- Definition
  - The ratio of the output in air for a given field to that for a reference field (10 x 10 cm)
- Direct measurement

$S_c = D(r)/ D(10)$

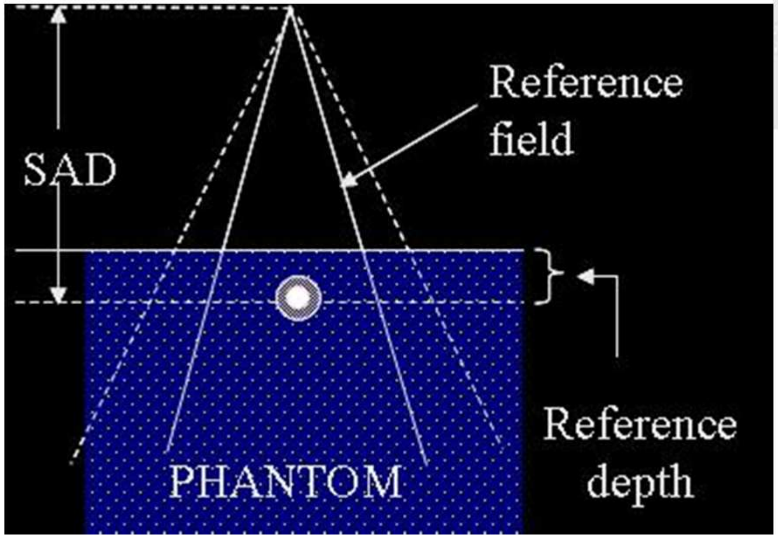


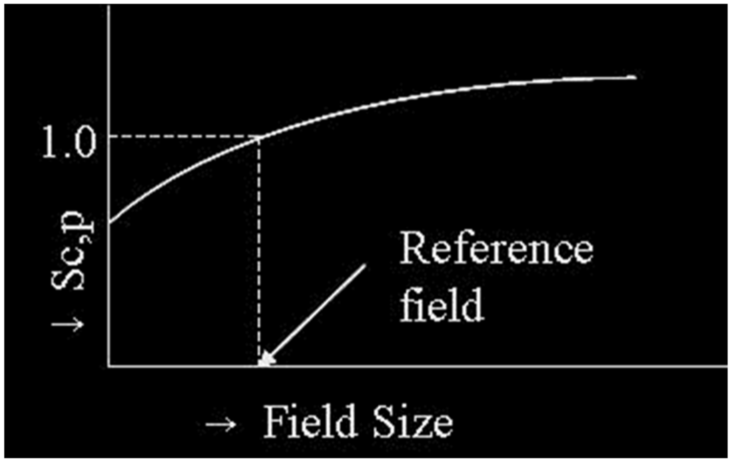


Phantom Scatter Factor (S_p)

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

$$S_p = \frac{S_{c,p}(r)}{S_c(r)}$$

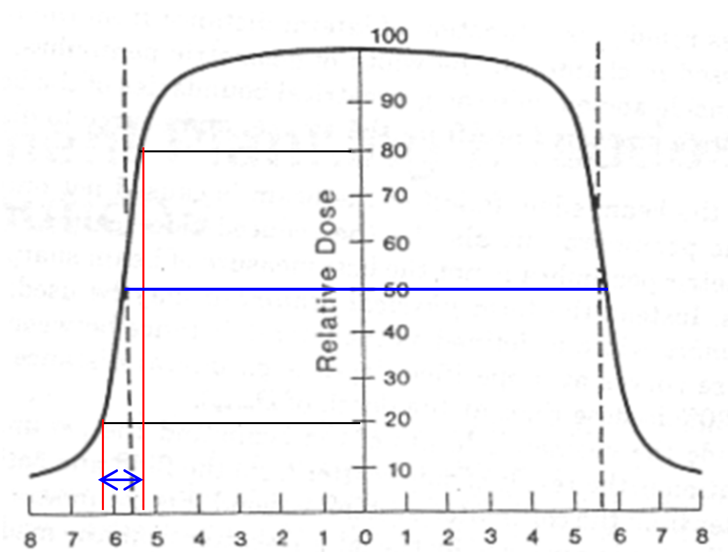


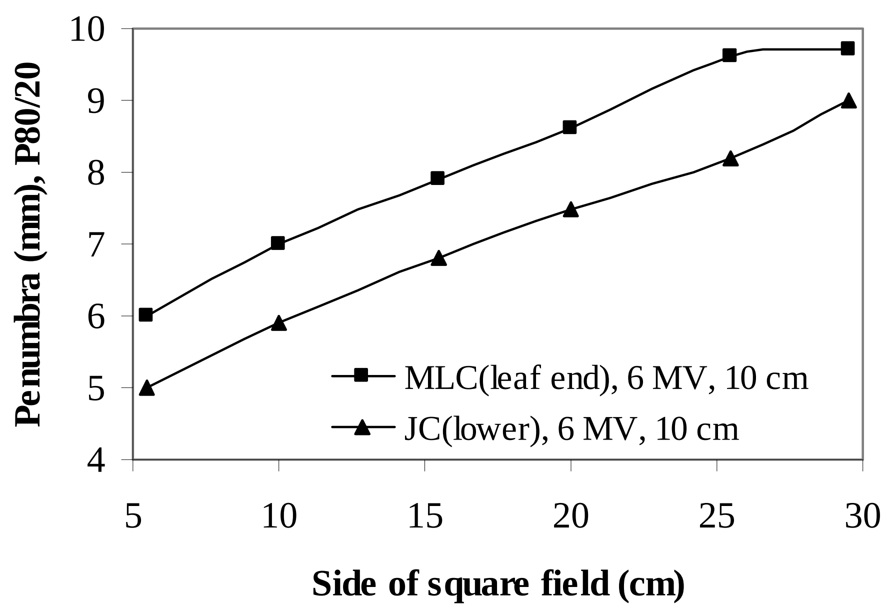


Photon Beam Penumbra

The penumbra region

- The dose rate decreases rapidly as a function of lateral distance from the beam axis.
- The width of geometric penumbra depends on source size, distance from the source, and source-to-diaphragm distance.



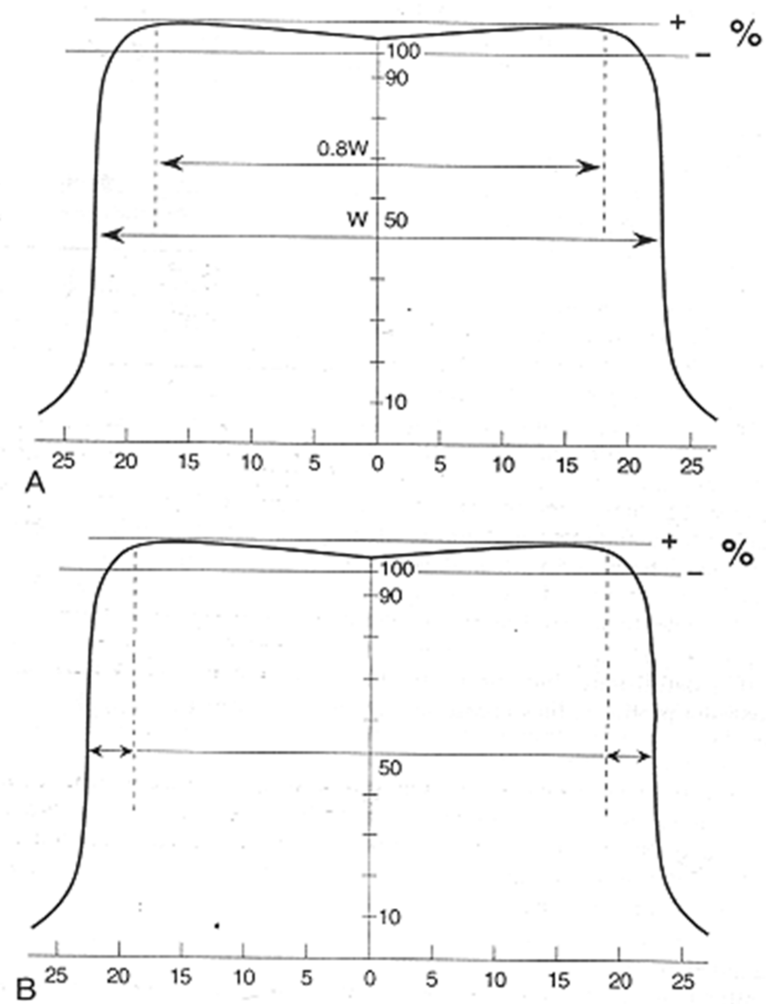


Flatness and Symmetry

- Flatness
  - within ±3%. over 80% of the field

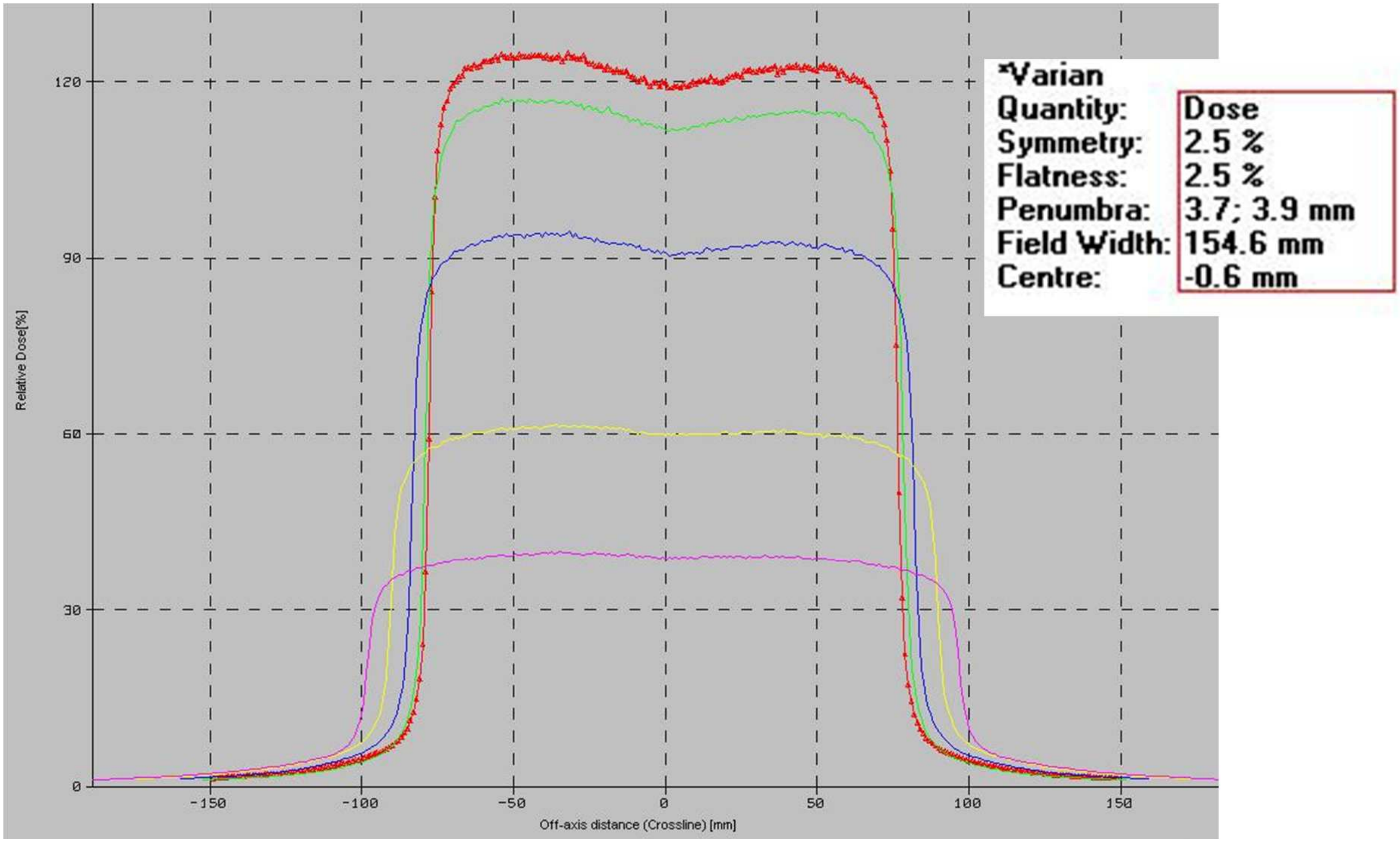
- Symmetry
  - within ± 2% over 80% of the field

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



Profile characteristics

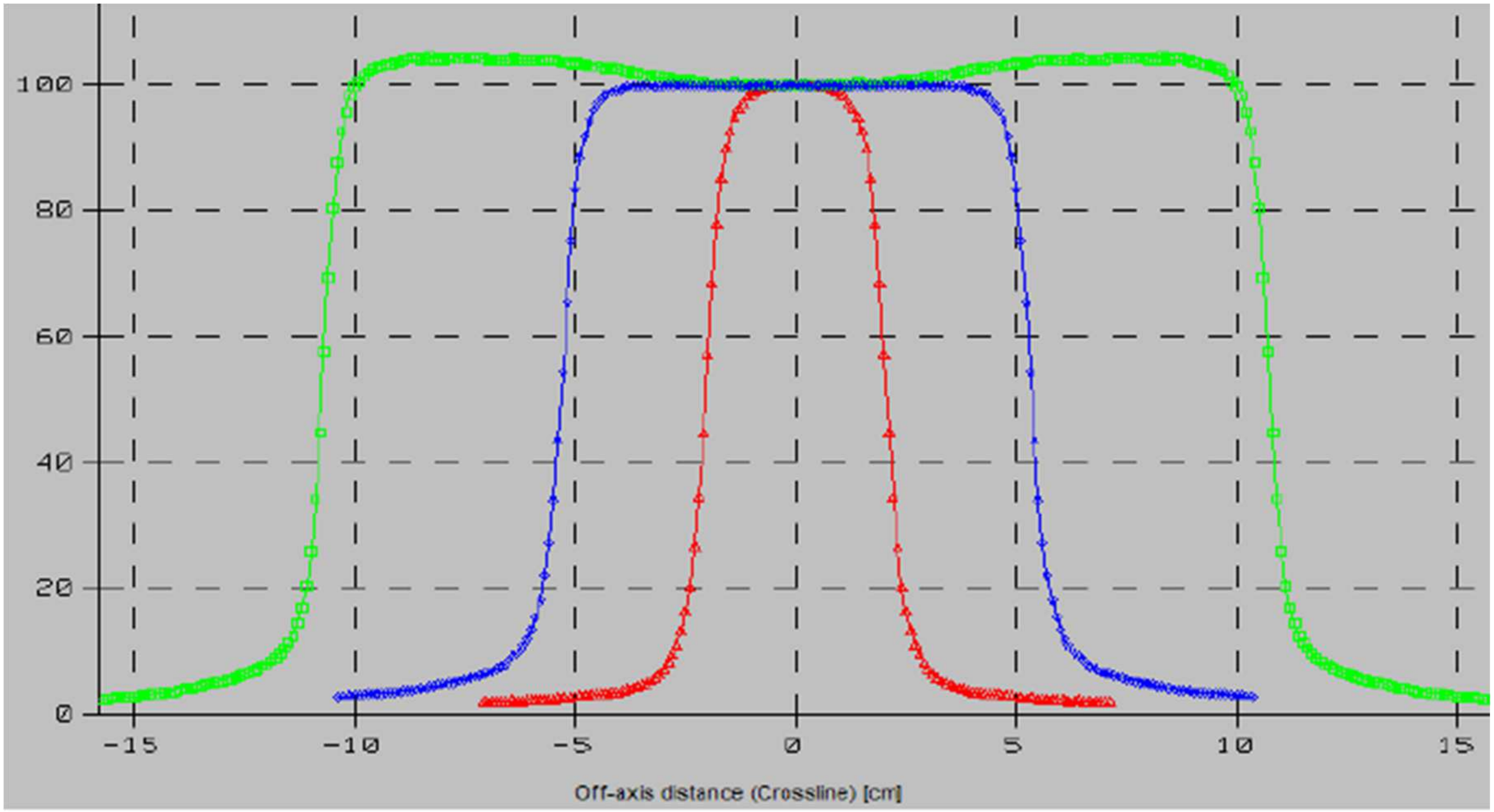
15 MV Photon Beam, Field size of 15x15cm^2, Depth 2.5, 5.0, 10, 15, 20 cm



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

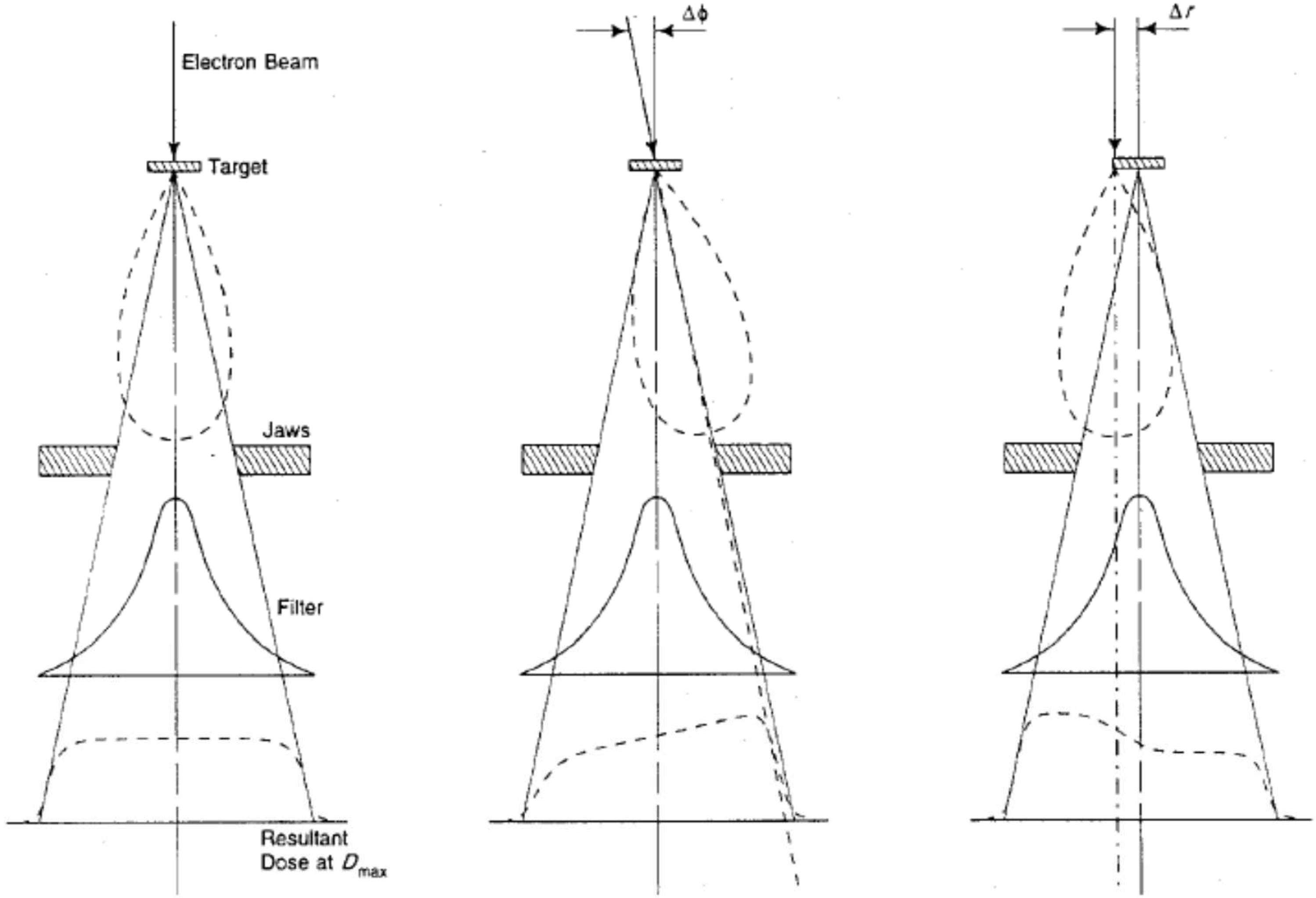
Cross Beam Profile

6 MV Photon Beam, Depth of 5.0 cm, Field size of 4x4, 10.4x10.4, and 21x21cm^2



The flatness of photon beams is extremely sensitive to change in energy of the incident beam. A small change in the penetrative quality of a photon beam results in very large change in beam flatness.

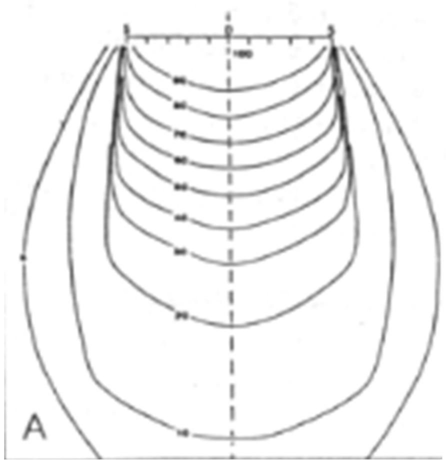
Effect of Electron Steering
Beam Flatness



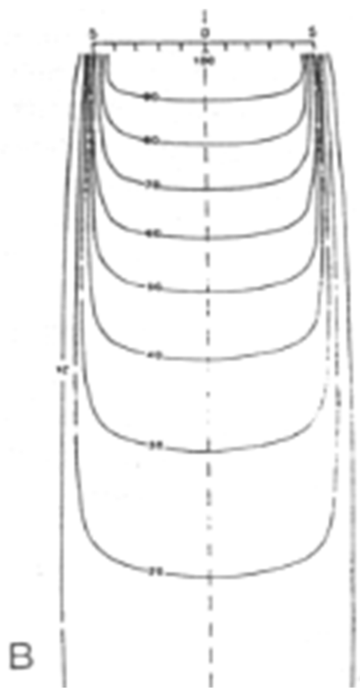
Symmetric Tilted Displaced

Beam Quality

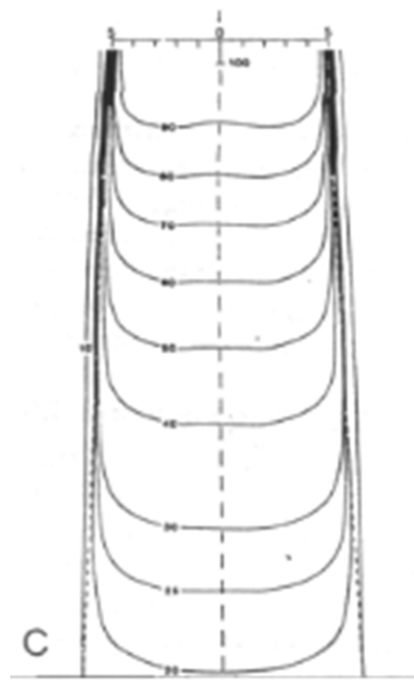
- The depth of a given isodose curve increases with beam quality.
- Greater lateral scatter associated with lower-energy beams
- For megavoltage beams, the scatter outside the field is minimized as a result of forward scattering and becomes more a function of collimation than energy.



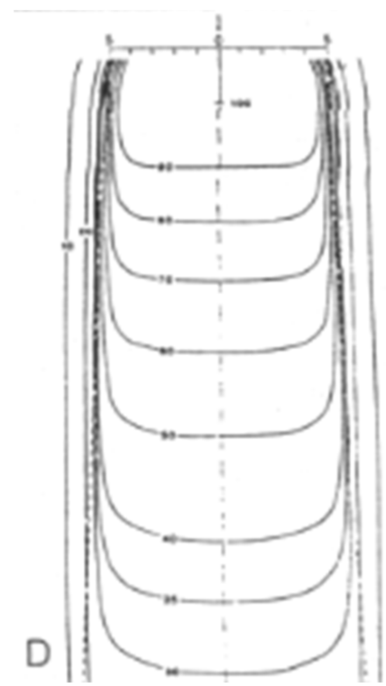
200 kVp, SSD=50 cm



^{60}Co, SSD=80 cm



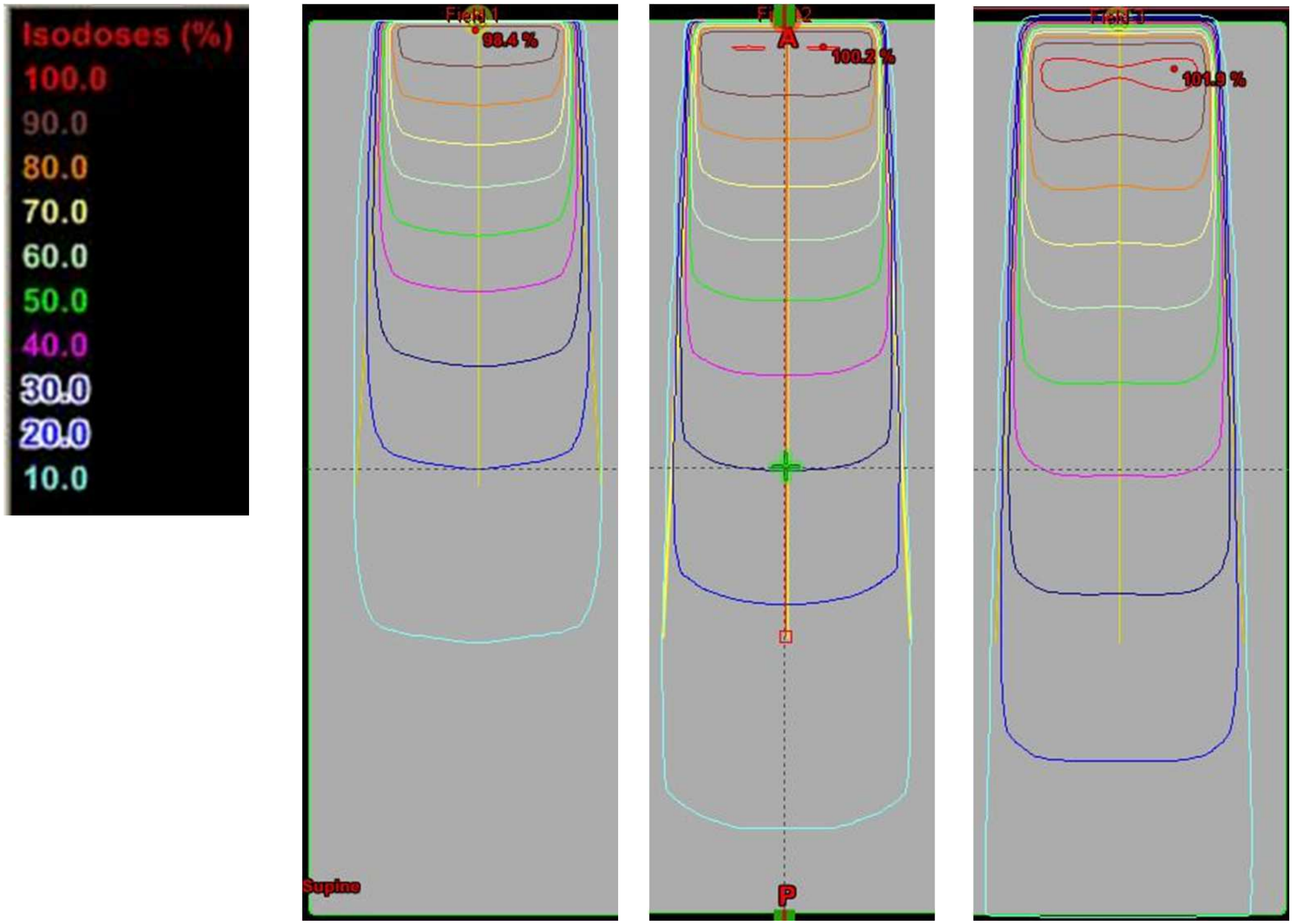
4 MV, SSD=100 cm



10 MV, SSD=100 cm

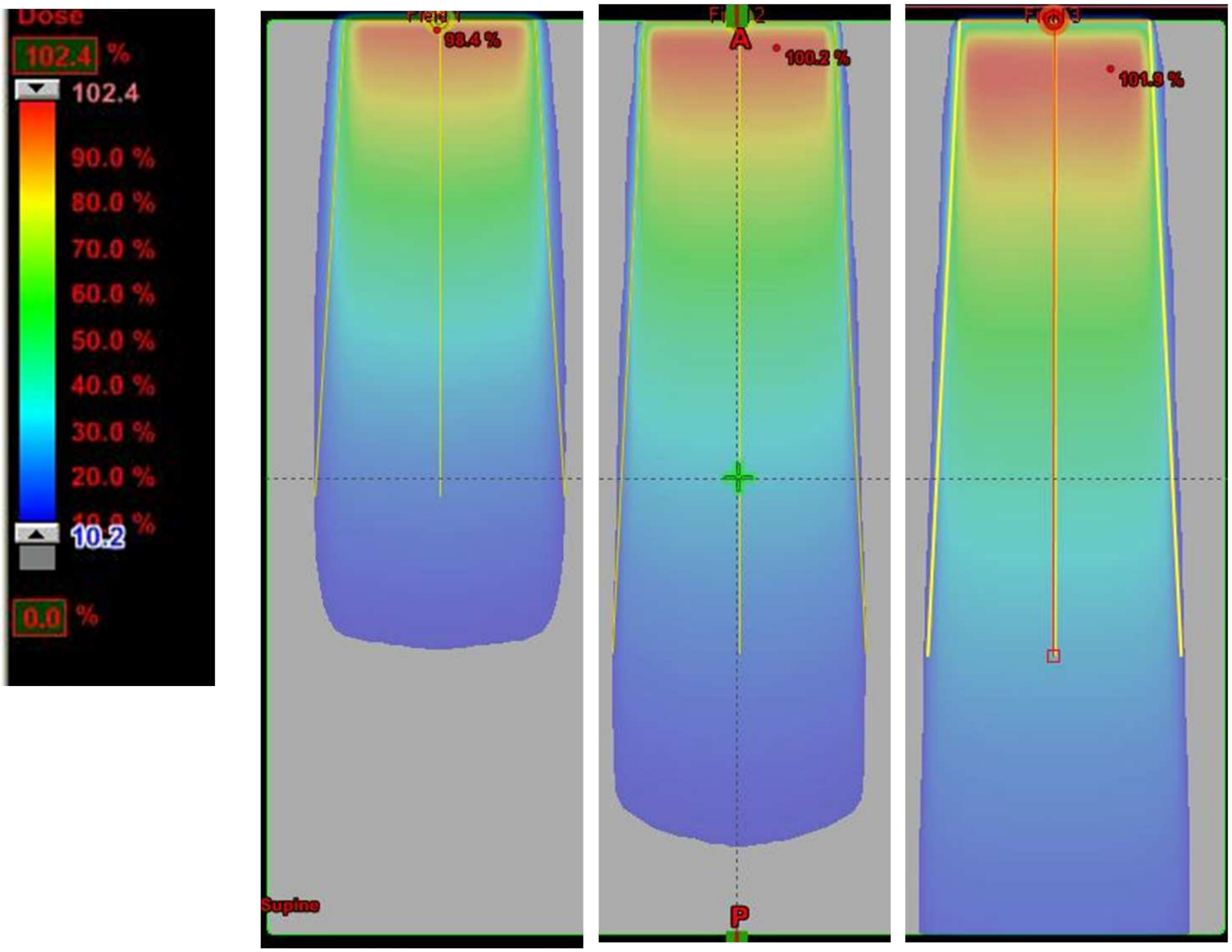
Isodose distribution

Co-60, 6 & 15 MV Photon Beam, Field size of 10x10cm^2



Isodose distribution

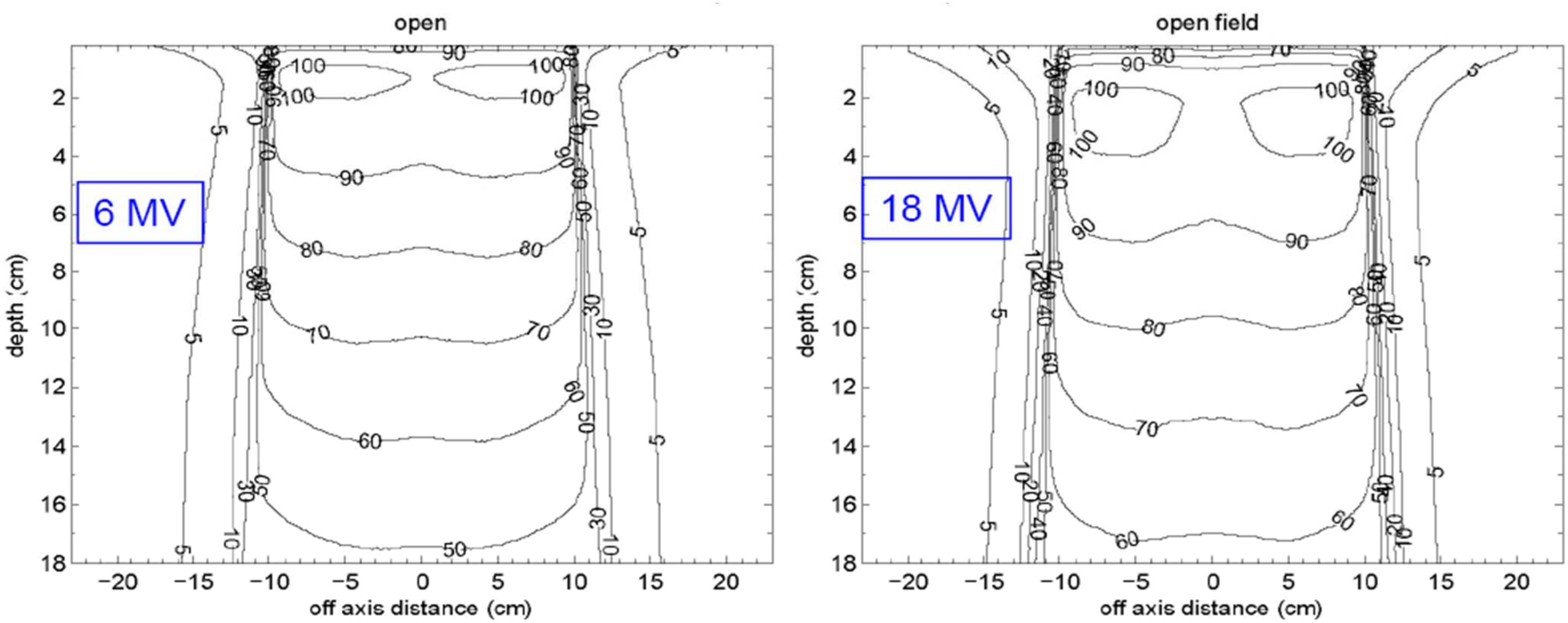
Co-60, 6 & 15 MV Photon Beam, Field size of 10x10cm^2



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

Isodose distribution

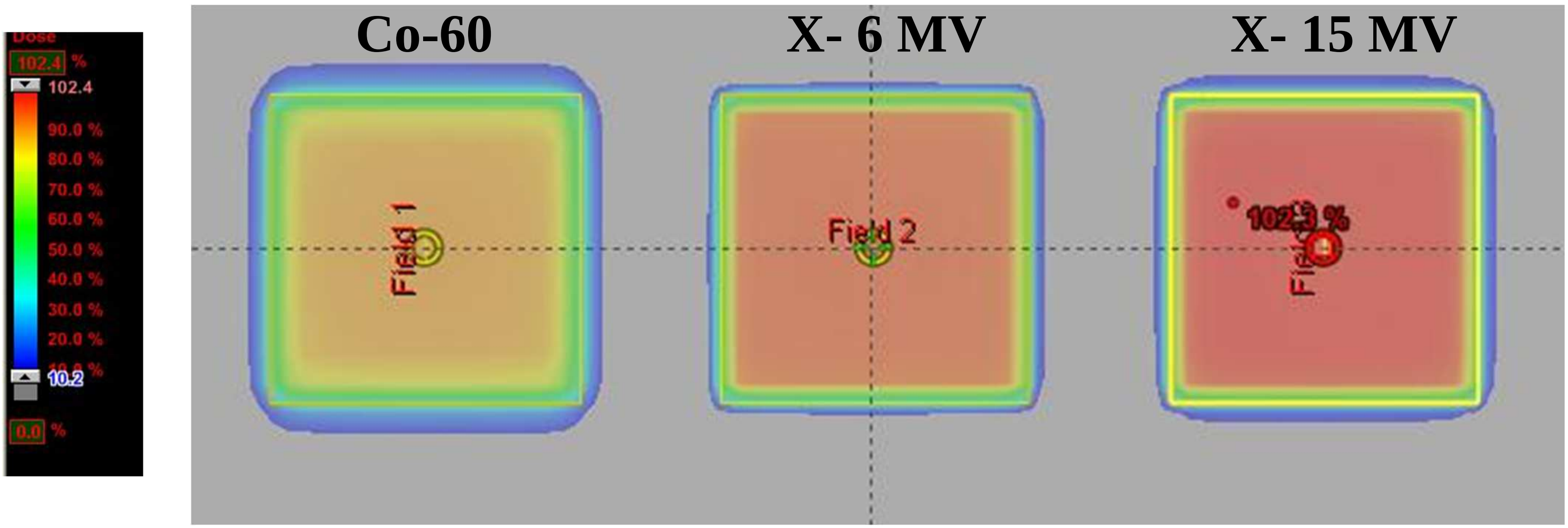
Field size of 20x20cm^2



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

Cross section isodose distribution

Co-60, 6 & 15 MV Photon Beam, Field size of 10x10cm^2



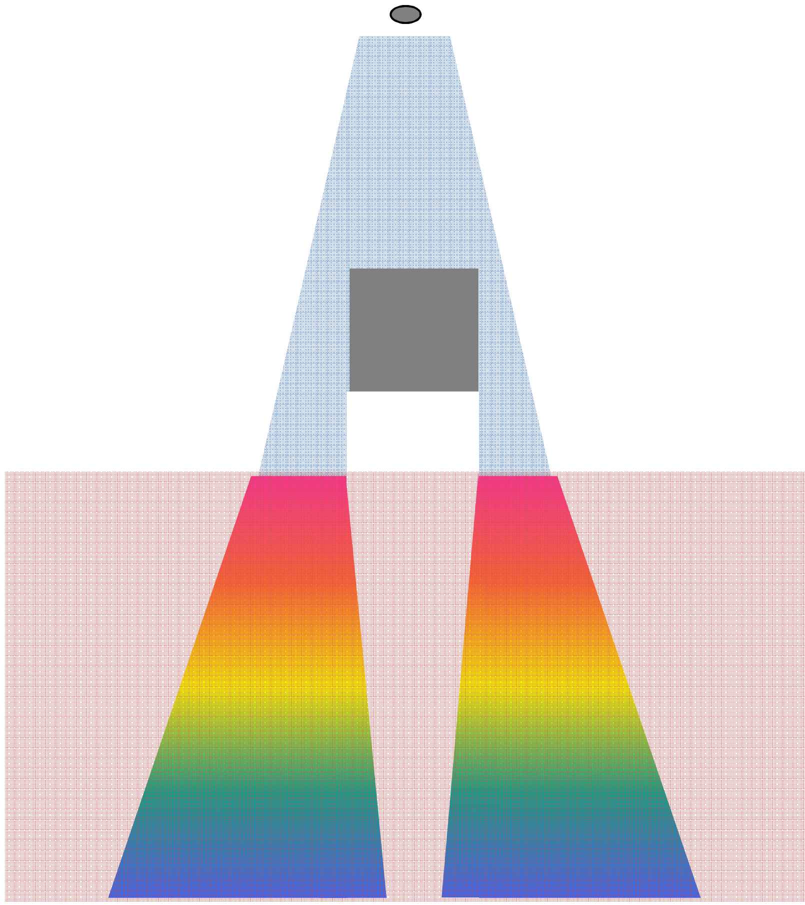
Problem in beam modification

- Radiation reaching any point, is made up of primary and scattered photons.

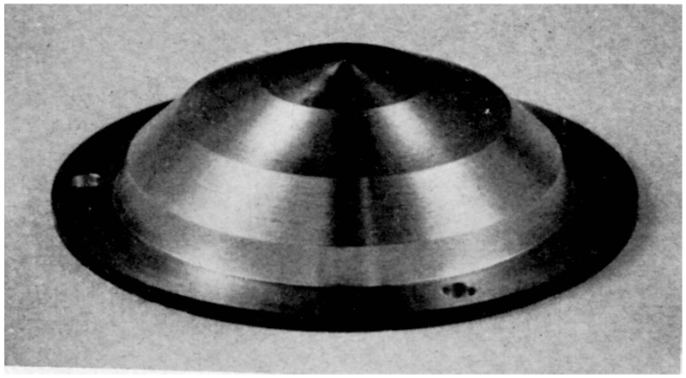
- Any introduction of the modification devices results in alteration of dose distribution, due to these two phenomena.

- The phenomena scattering results in an "blurring" of the effect of the beam modification.

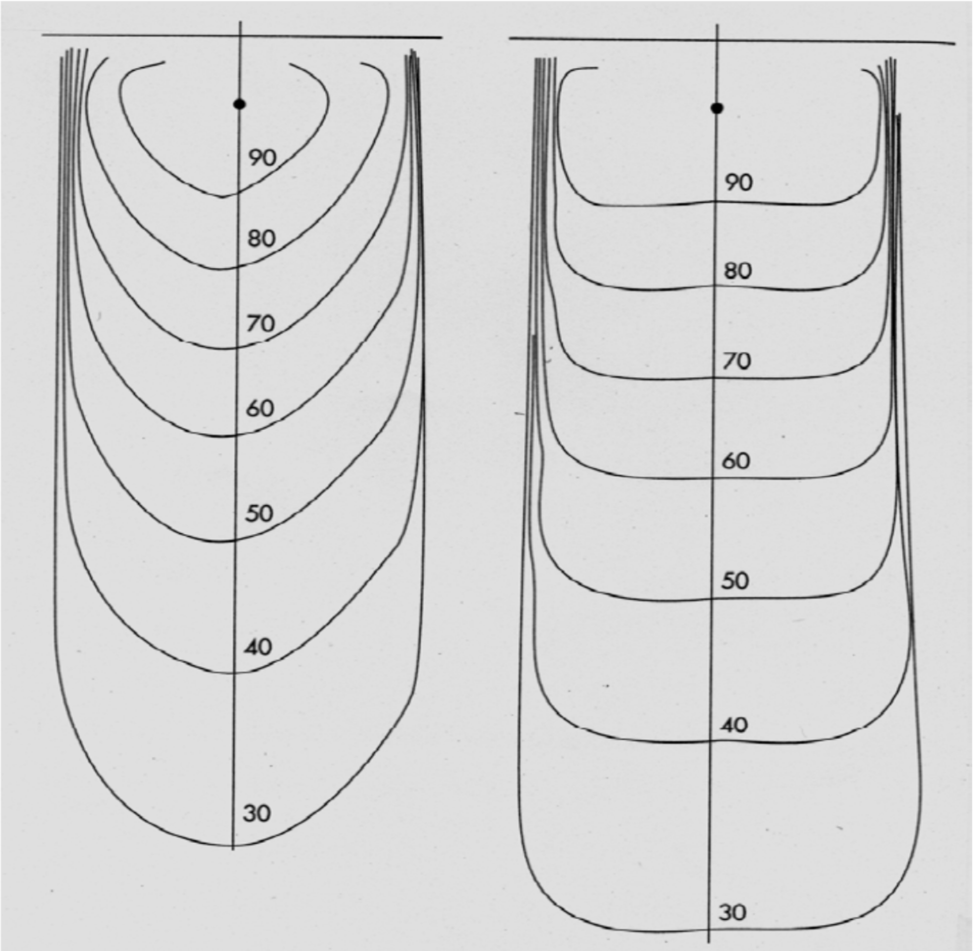
- Scattering is more in kilovoltage radiation than in megavoltage radiation therapy.



Beam flattening filter

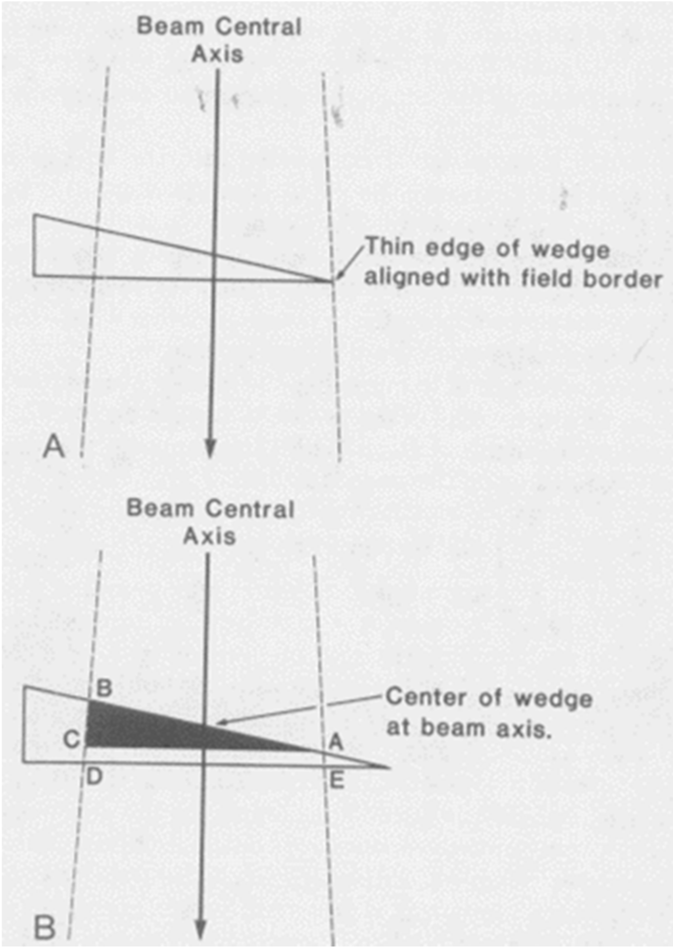


- Isodose curve for a 10 MV x-ray beam without (Left) and with (right) beam-flattening filter in place.
- Lateral horns of the curves are apparent near the surface with the beam-flattening filter.
- For IMRT purposes we explore using FFF beam



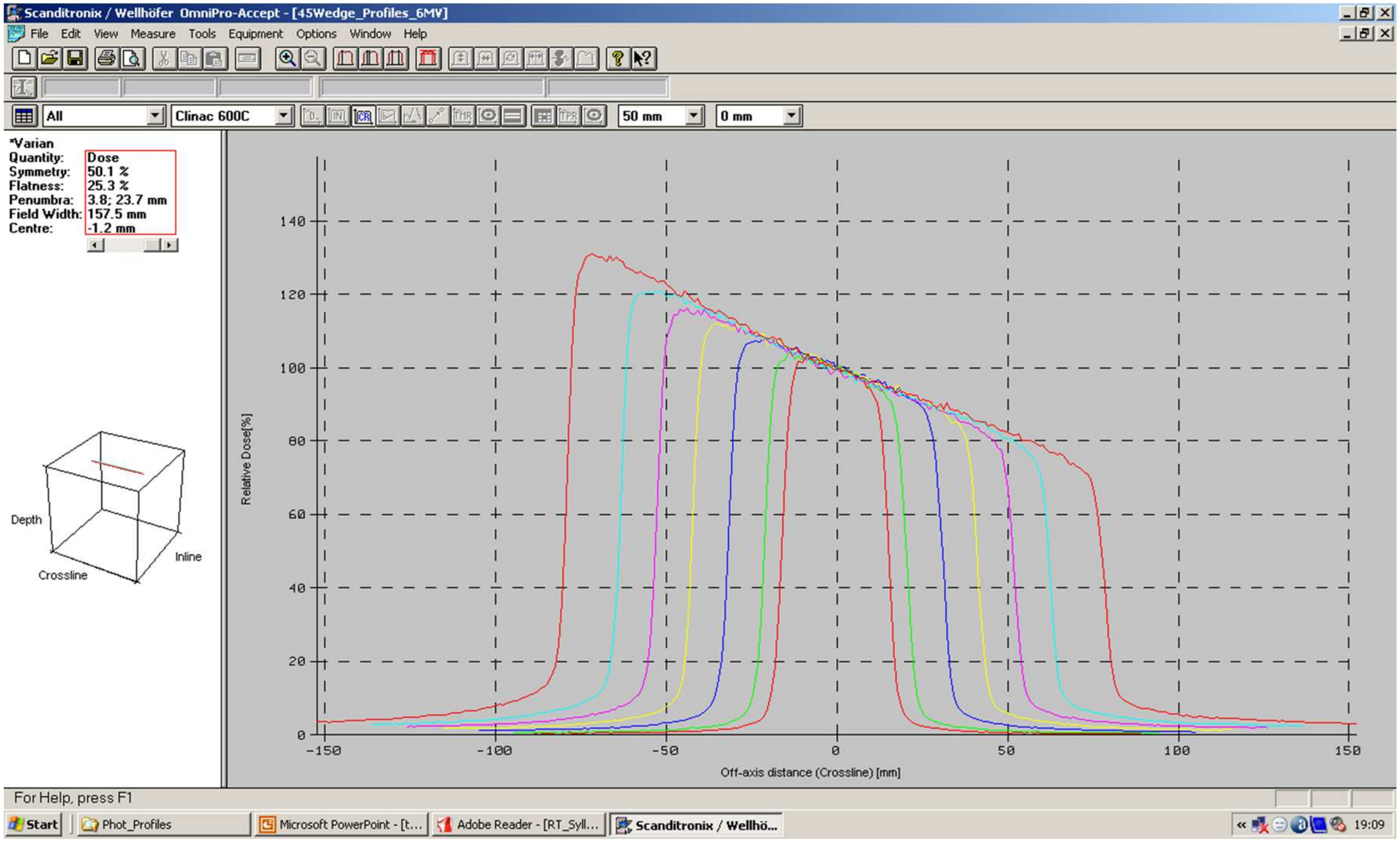
Wedge Filter

- Wedge shaped absorber which causes a progressive decrease in the intensity across the beam, resulting in a tilt of the isodose curves from their normal position.
- Made of dense material : Lead, copper or steel
- Wedge transmission factor, < 1

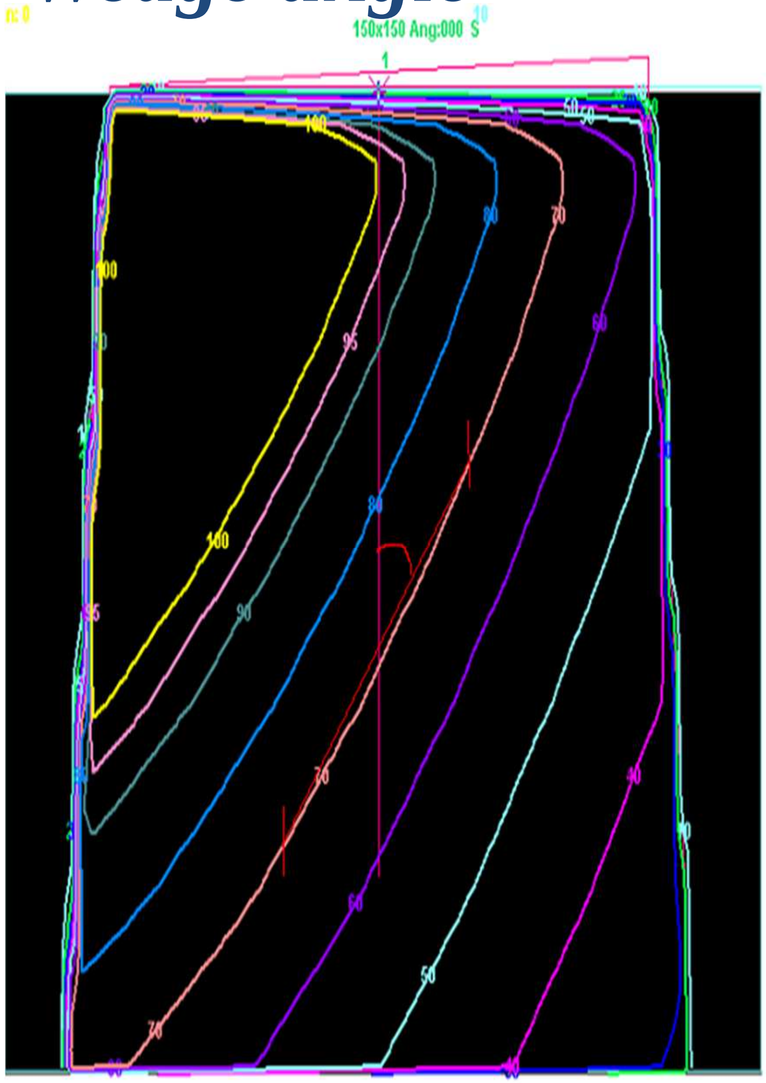


- Individualized wedge system
  - A separate wedge for each beam width
  - to minimize the loss of beam output
  - To align the thin end of the wedge with the border of the light field
  - Used in ^{60}Co
- Universal wedge system
  - A single wedge for all beam widths
  - Fixed centrally in the beam
  - Used in Linac

Wedge profile at 5cm depth (45 degree)



Wedge angle



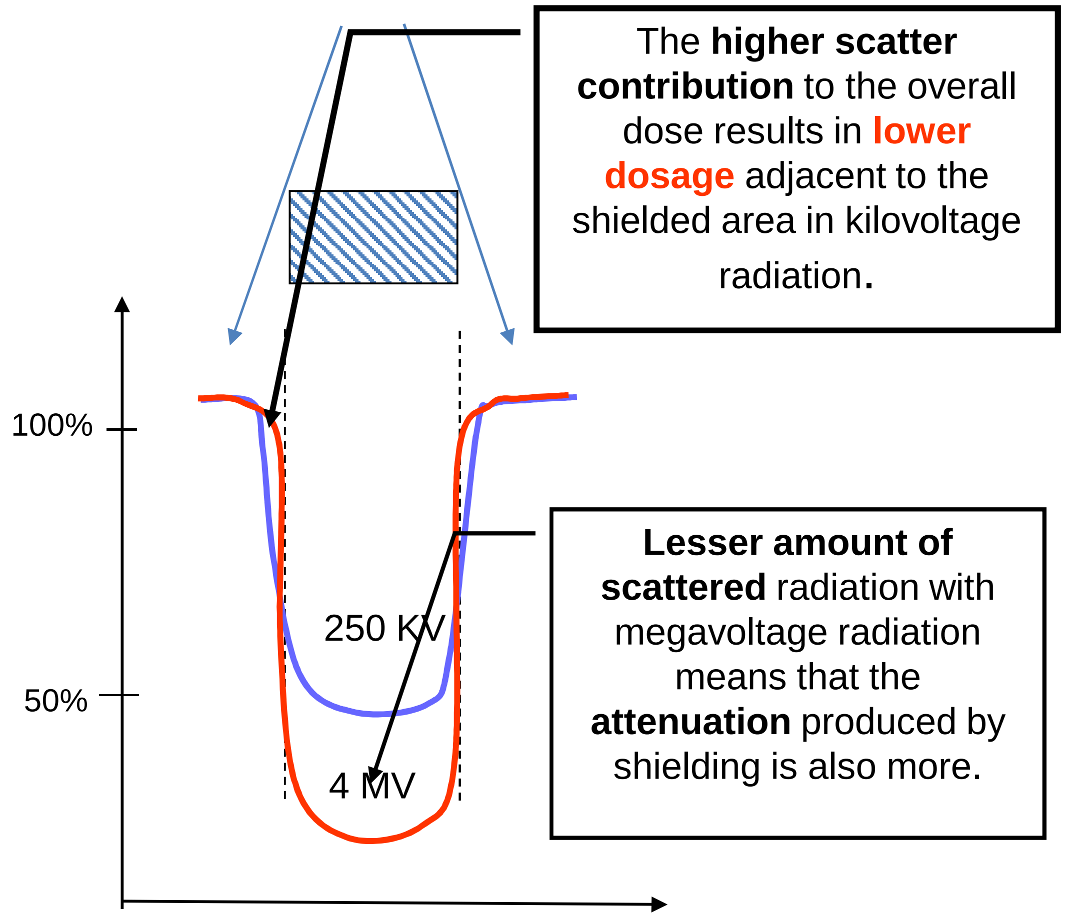
- The wedge isodose angle (θ) is the **complement** of the angle through which the isodose curve is tilted with respect to the **central ray** of the beam at any specified depth.
- This depth is important because the angle will **decrease** with increasing depth.
- The choice of the reference depth varies:
  - **10 centimeters.**
  - 1/2 - 2/3rd of the beam width.
  - At the 50% isodose curve (kV).

It is angle is defined as the complement of the angle through which the isodose curve with respect to the beam central axis at reference depth of 10cm.

(ICRU Report N0.24)

Beam modifiers

- Field blocking and shaping devices:
  - Shielding blocks.
  - Custom blocks.
  - Asymmetrical jaws.
  - Multileaf collimators.



Shielding blocks

- To spare the critical organ & Normal tissue
- Should be at least 5 HVL (3.125%) ; $1/2^n$ = % transmission
- Made of

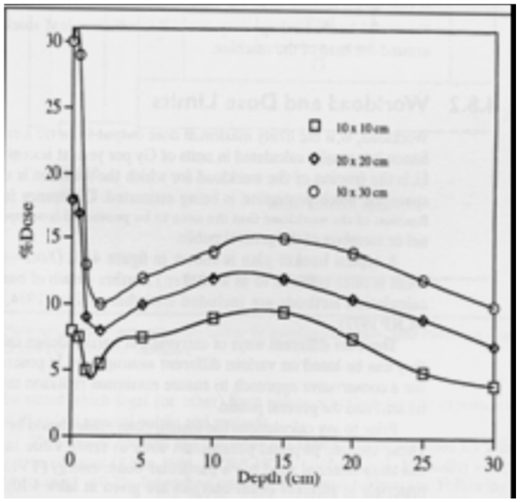
  Lead 11.3gm/cm^3

  Cerro-bend alloy 9.4gm/cm^3

  (Bi –50%, Pb –26.7%, tin –13.3% & Cd -10%)

Beam Quality	5 HVL Lead (cm)
Cs –137	3.0
Co – 60	5.0
4 MV	6.0
6 MV	6.5
10 MV	7.0
15 MV	7.0

Dose under blocks: %DD under a 2 cm wide block

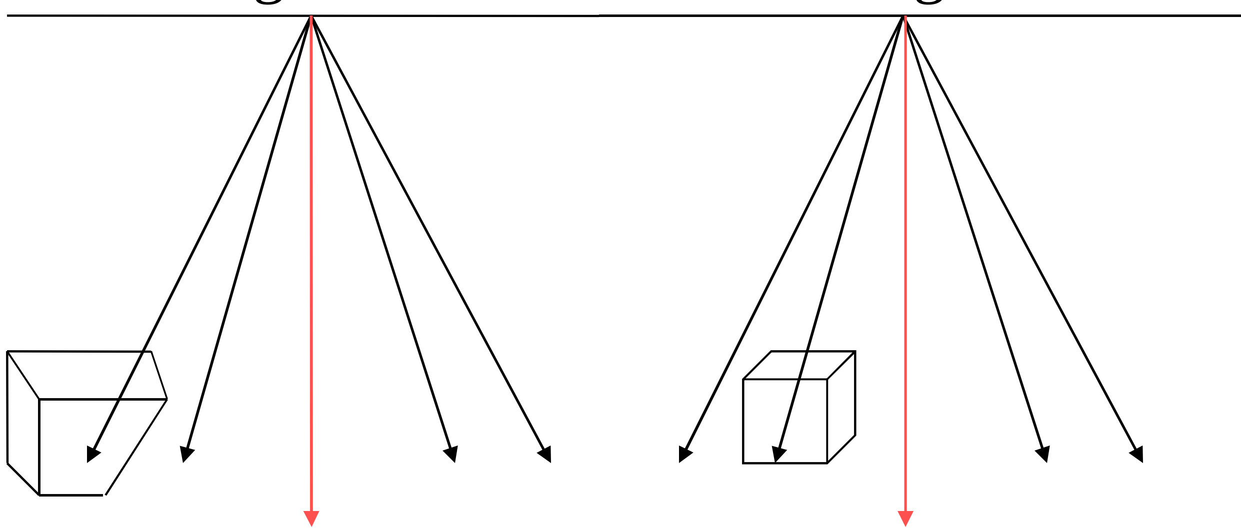


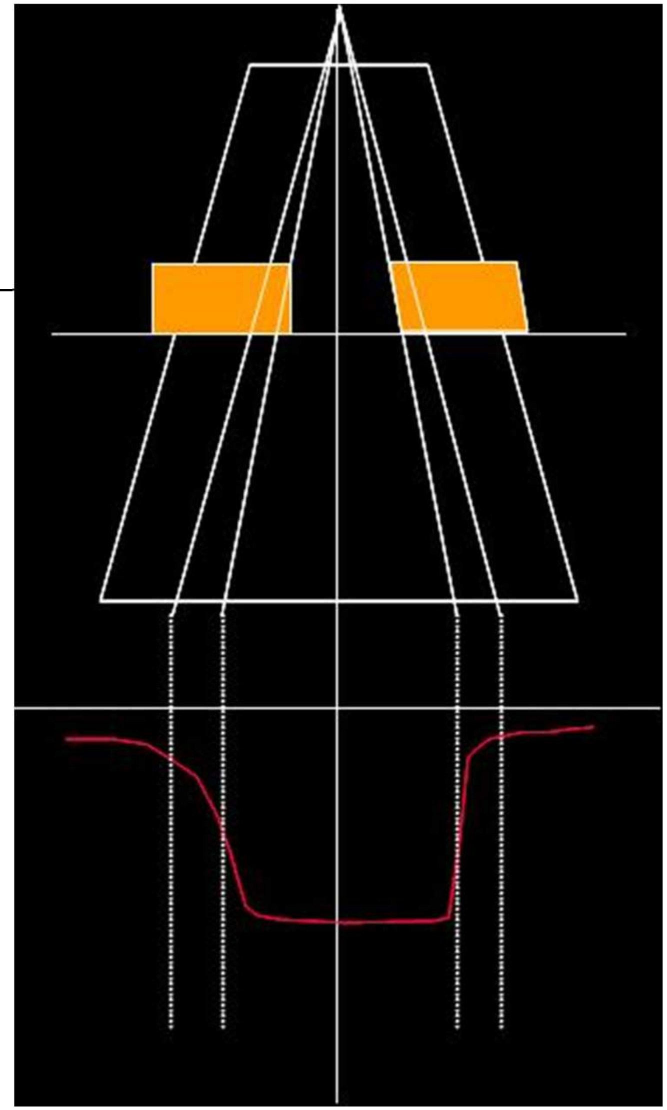
Larger fields produce more dose under the block due to increase in tissue scatter

Note: initial dose is high due to electron contamination, followed by a rapid reduction in dose, then a slow climb to a plateau at about d=15cm

Regular Vs Divergent shielding block

Divergent blocks Non-divergent blocks

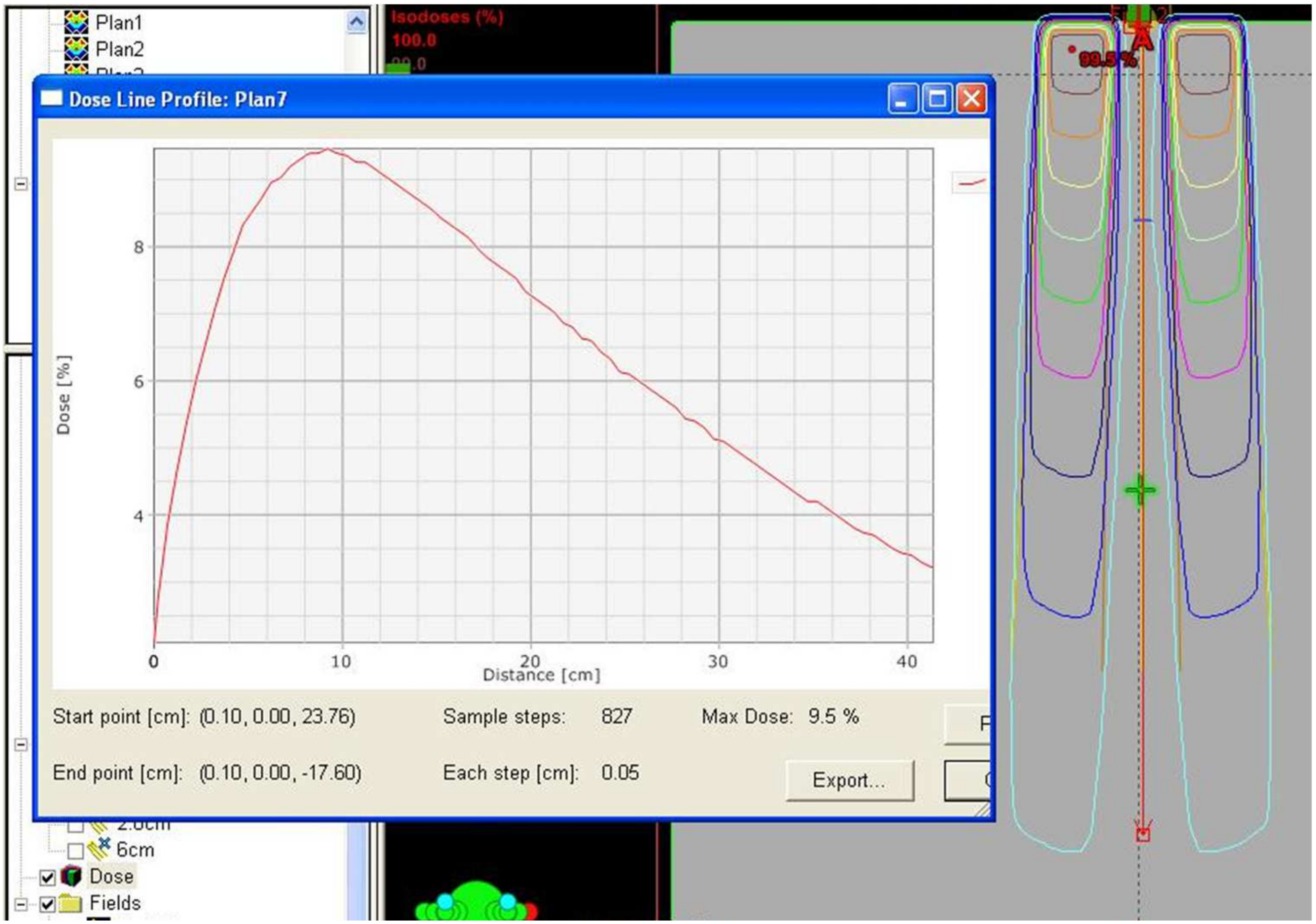


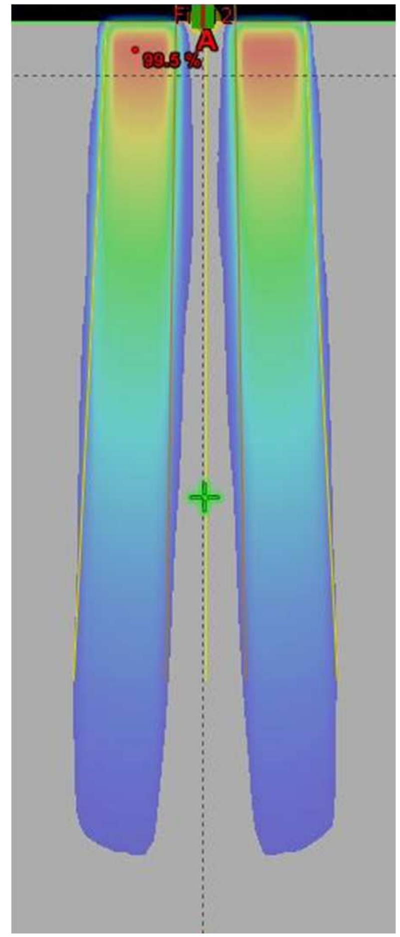


- Sharp penumbra
- Tighter shielding margin
- Particular STD

- Increased penumbra
- Larger shielding margin
- Any STD (20cm clearance)

Isodose distribution with shielding block





Concepts in treatment planning

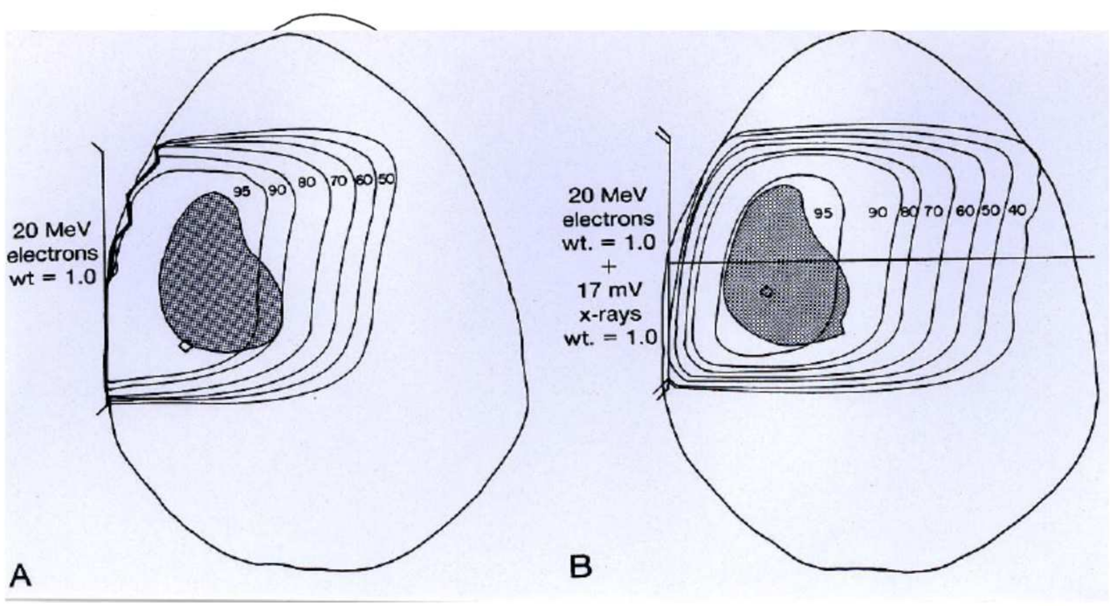
- Beam arrangement
- Beam weighting
- Fixed treatment technique
- Isocentric treatment technique
  - Co planner & non- co planner
- Beam blocking
- Asymmetric collimation
- Intensity modulation

Criteria for Using Single Enface Treatment Fields

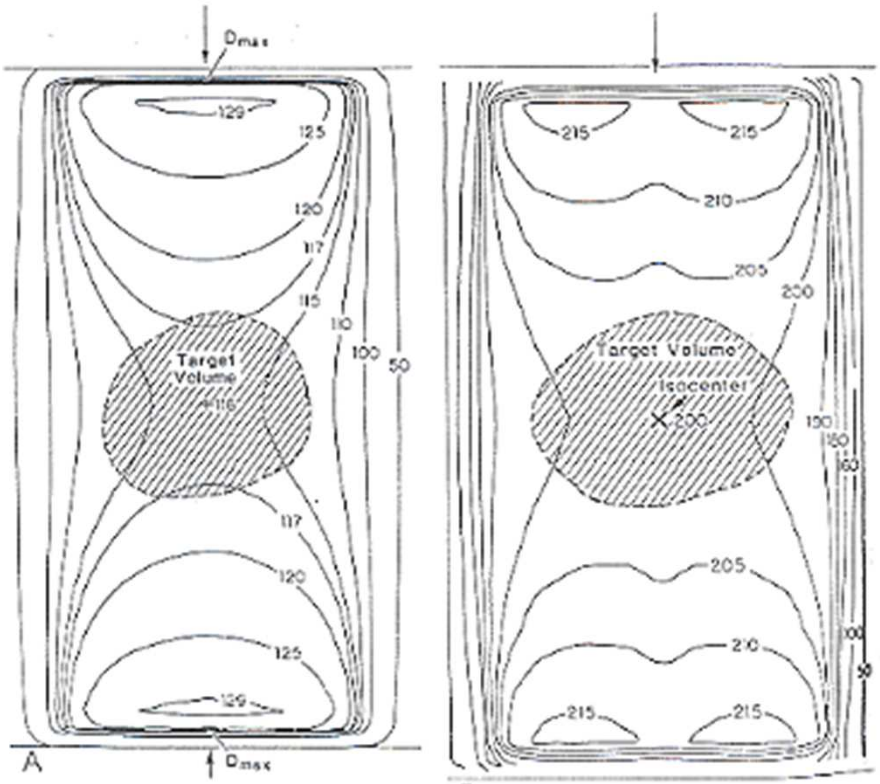
- Dose distribution within the tumor volume is reasonably uniform (+ 5%)
- Maximum dose is not excessive, not more than 110% of prescribed dose
- Critical structures are kept below tolerances

Examples of enface fields:

- a) s’clav
- b) internal mammary
- c) spinal cord compression



Parallel Opposed Fields



A, Each beam weighted 100 at D_{max}.

B, Each beam weighted 100 at the isocenter.

The advantages

- The simplicity and reproducibility of setup
- Homogeneous dose to the tumor
- Less chances of geometrical miss

Disadvantage

- The excessive dose to normal tissues and critical organs above and below the tumor

Characteristics of parallel opposed fields are as follows:

- Hour glass shape of the 100% isodose curve
- A uniform distribution at the patient midline

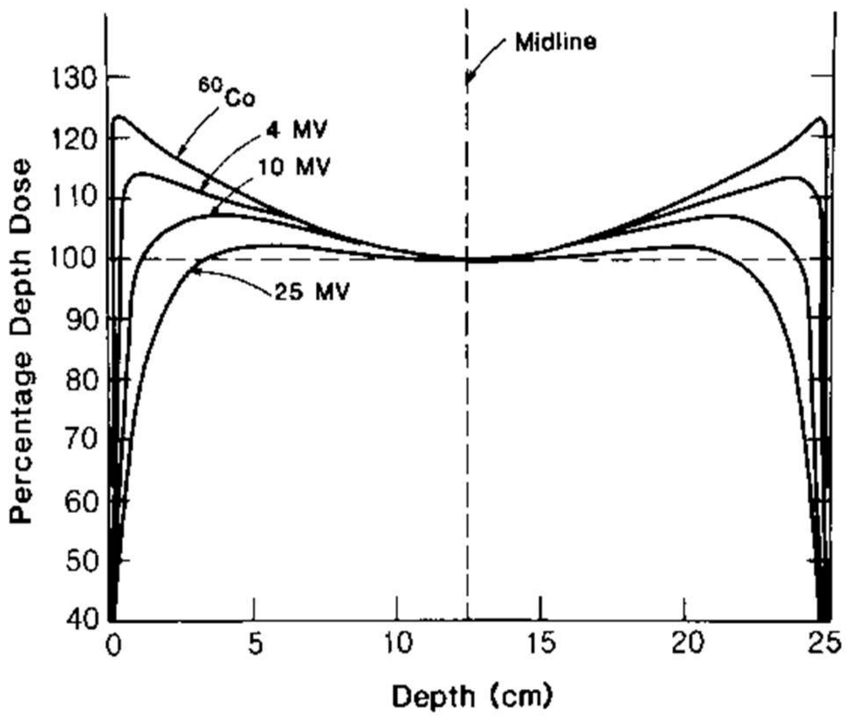
Patient Thickness vs Dose Uniformity

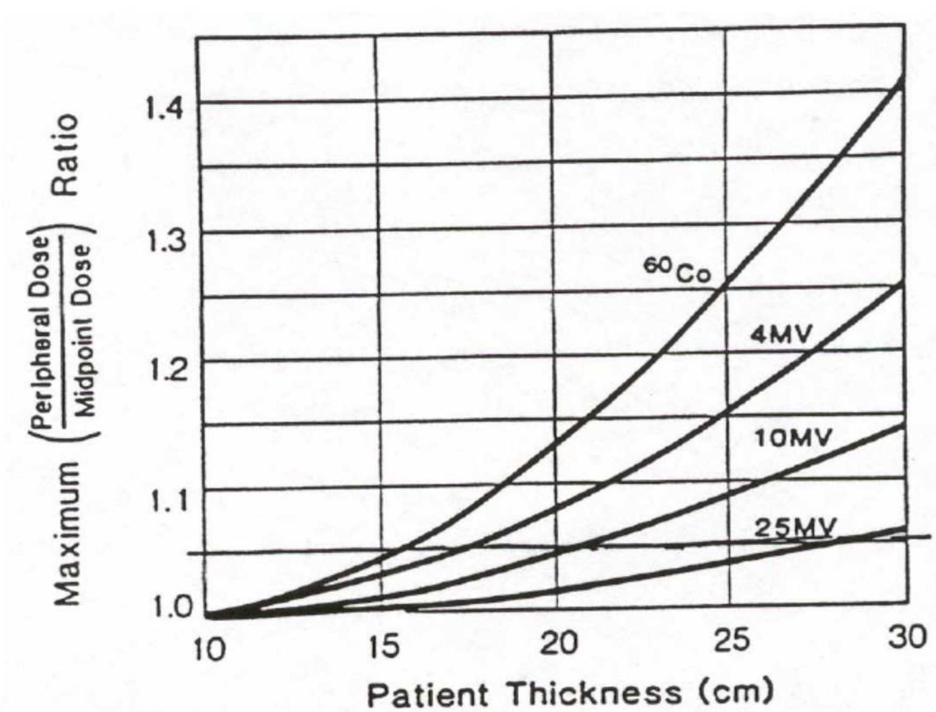
Parallel opposed beams give a uniform dose distribution across the patient.

Dose uniformity depends on thickness, energy, and beam flatness

Dmax dose increases as either

- thickness increases
- energy decreases



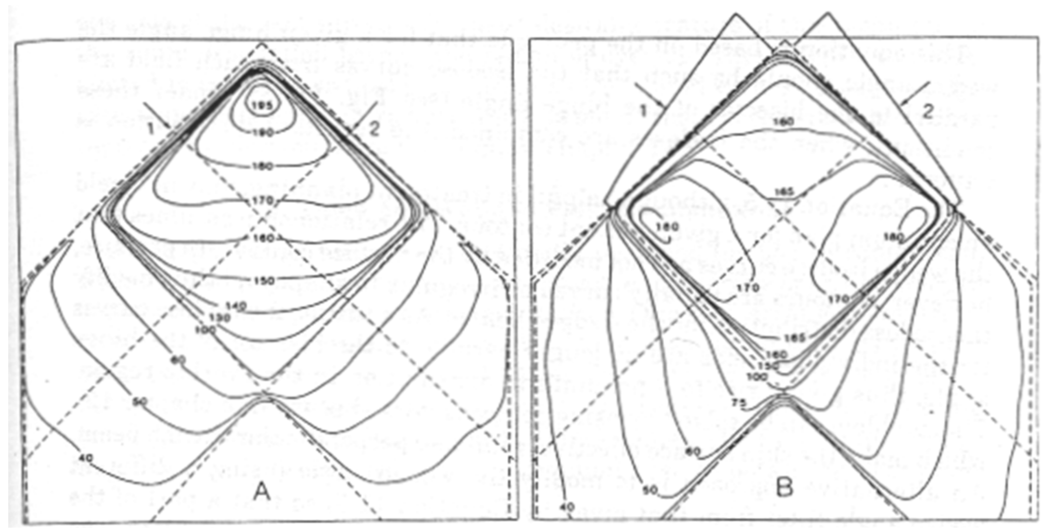


Isocentric Techniques

- The isocenter is the point of intersection of the collimator axis and the gantry axis of rotation.
- Isocentric technique
  - Placing the isocenter at a depth with the patient and directing the beams from different directions
  - SSD = SAD-d
- Stationary beams
- Arc & rotational beams

Wedge Field Techniques

- The dose gradient in the overlap region is minimized.
- The dose falls off rapidly beyond the region of overlap or the “plateau” region.



Wedge angle θ = 90º- ϕ/2

where ϕ = the hinge angle

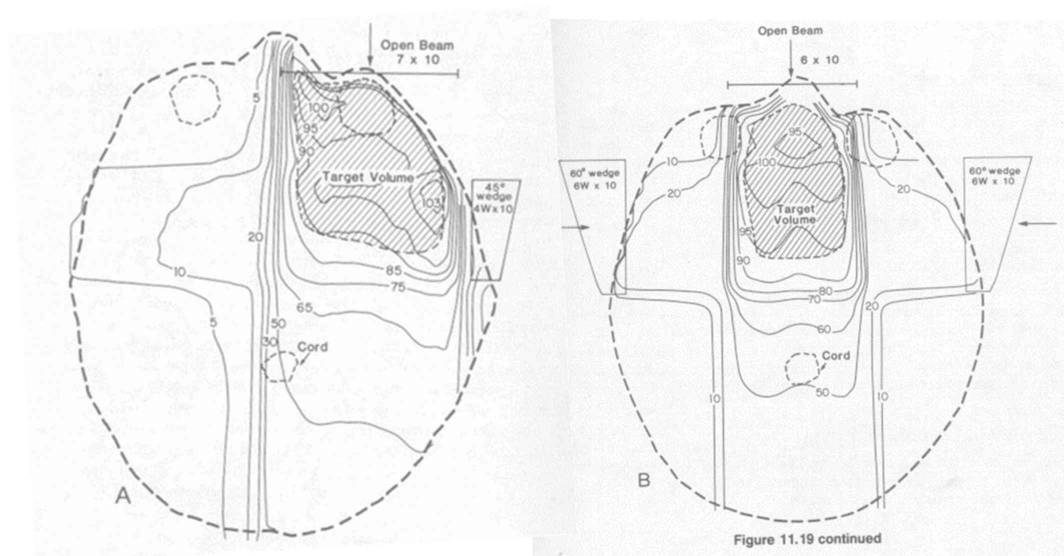
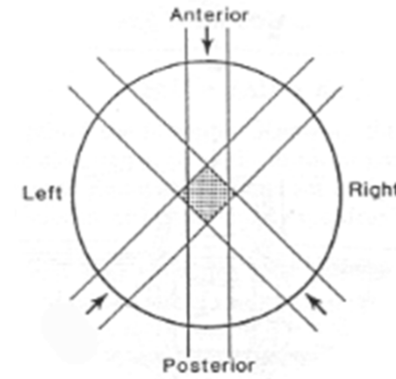


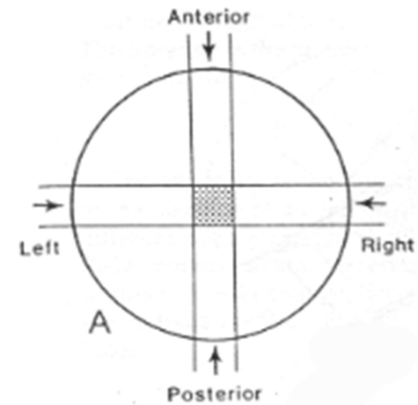
Figure 11.19 continued

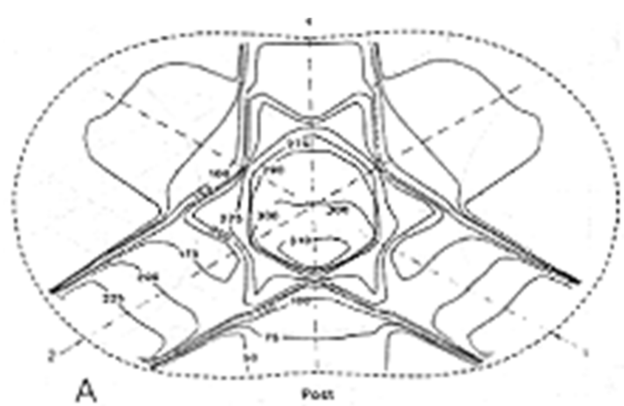
Multiple Fields

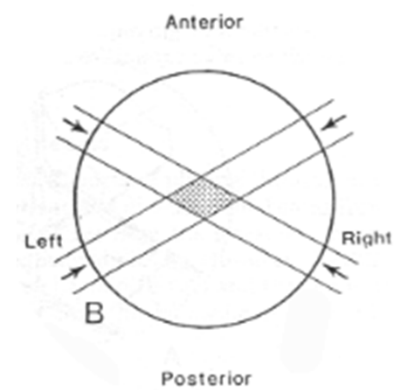


To deliver maximum dose to the tumor and minimum dose to the surrounding tissues

- Using fields of appropriate size
- Increasing the number of fields or portals
- Selecting appropriate beam directions
- Adjusting beam weights
- Using appropriate beam energy
- Using beam modifiers







Summary

- **Basic properties of photon beams :**

  Quantity, Quality, Intensity, Linear attenuation & HVL

- **Parameters that influence the beam profile characteristics** :

  PDD, TMR, Buildup, Scatter, Field size & SSD dependence & Penumbra

- **Influence of beam modifiers:**

  Flattening filter, Wedge & Shielding blocks

- **Dose distribution:**

  Energy dependence, Penumbra & Contaminant electrons

- **Basic concepts of treatment planning :**

  Single versus multiple beams and techniques

Thank you for your attention!!