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Monitor unit calculation for photon beams: theory

ICTP School of Medical Physics for Radiation Therapy: Dosimetry and Treatment Planning for Basic and Advanced Applications

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The Abdus Salam International Centre for Theoretical Physics
IAEA International Atomic Energy Agency
United Nations Educational, Scientific and Cultural Organization

Introduction
MU computation in RT

- Application of dosimetry concept and quantities for dealing with dose computation problems in clinical practice
- Data extracted from phantom measurements are used
- Specific problem: calculate the MU on the treatment unit wich will deliver an intended dose

Introduction

Problem:

To compute the dose (or MU) in any point in a phantom (patient) for any field

By knowning the dose at the Normalization Point for a standard field determined during calibration procedures

Introduction

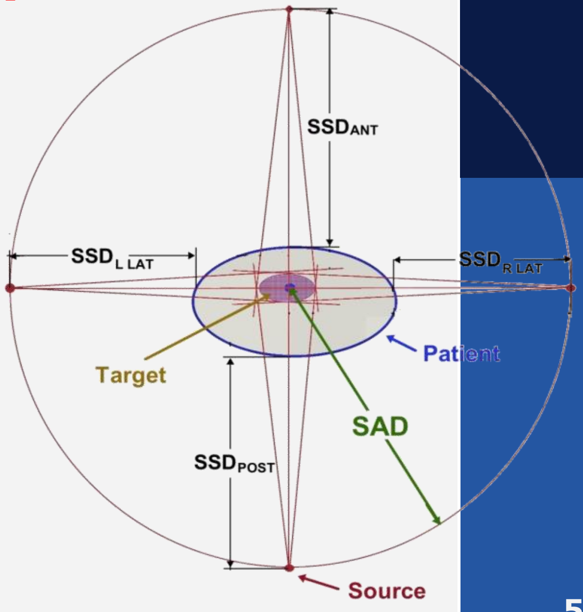
The dosimetric functions used for dose computation are based on water phantom measurement:

1. **Standard SSD**
2. **Perpendicular incidence**
3. **Ionization chamber or solid state devices are used**

Introduction
Set-up considerations

- If a **single beam** is normally used for treatment it is natural to mantain the SSD at nominal fixed value → **SSD setup**

- Modern radiotherapy use **multiple beams** with different **beam incidence angles** much pratical to mantain a fixed patient setup and rotate the gantry → **SAD setup**



The AAPM TG-71 report formalism

- A protocol is presented for the calculation of MU, for constant source-surface distance (SSD) and source-axis distance (SAD) setups.
- The protocol defines the nomenclature for the dosimetric quantities used in these calculations, along with instructions for their determination and measurement.
- For photon beams, this task group recommends that a normalization depth of 10 cm be selected
- If the normalization point is at isocenter, the computation simplify

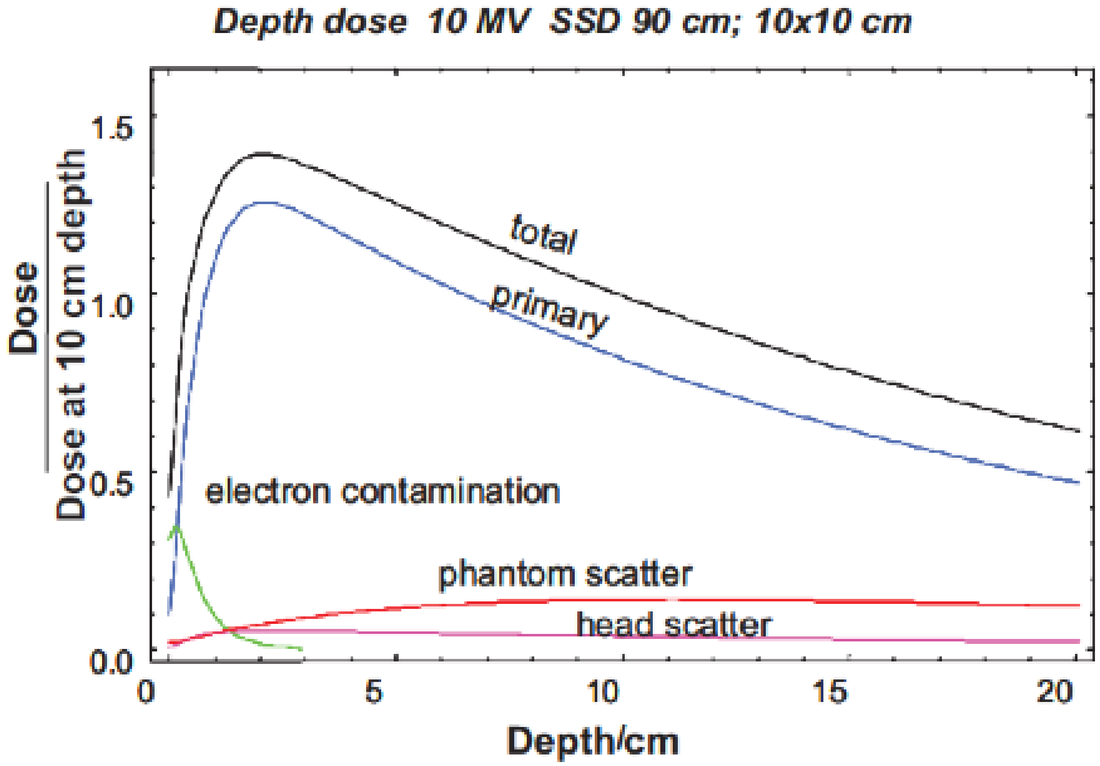


Figure 5.9 Depth dose distributions for a 10×10cm² field of 10 MV photons, showing separately the direct beam primary dose (blue), direct beam phantom scatter dose (red), electron contamination (green), total head scatter dose (pink) and the total sum of all components (black). Normalization is versus the total dose at the calibration position and field, which is the preferred normalization for comparing calculated and measured dose data.

Correction factors

The MU at prescription point are the result of the application of moltiplicative "correction factors" dependent by:

- **Depth** of the prescription point
- **Distance** from the X rays source
- **Size** of the beam
- Beam **modifiers** such as wedge filters or tray
- **Calibration** factor (recommended to set to 1)

The Monitor Unit Equation
Photon calculation using Tissue Phantom Ratio

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d, r_d) \cdot \mathrm{WF}(d, r_d, x) \cdot \mathrm{TF} \cdot \mathrm{OAR}(d, x) \cdot \left(\frac{\mathrm{SSD}_0 + d_0}{\mathrm{SPD}}\right)^2}.$$

In the case where dose is calculated at the isocenter point, Eq. (1) reduces to

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d, r_d) \cdot \mathrm{WF}(d, r_d) \cdot \mathrm{TF} \cdot \left(\frac{\mathrm{SSD}_0 + d_0}{\mathrm{SAD}}\right)^2}.$$

The Monitor Unit Equation

Photon calculation using percentual depth dose

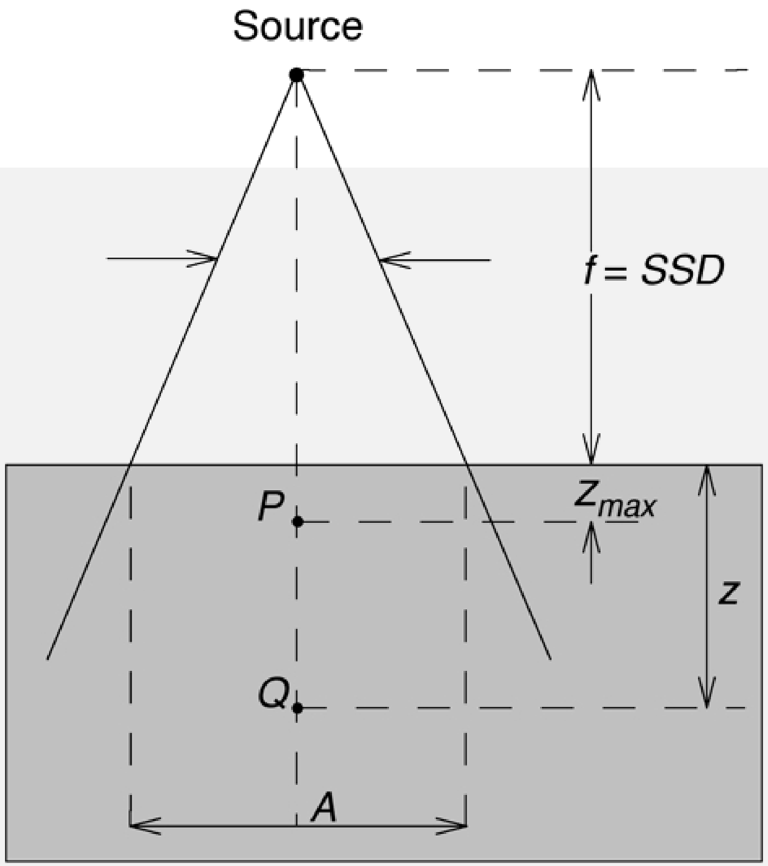
$$\mathrm{MU} = \frac{D \cdot 100\%}{D_0' \cdot S_c(r_c) \cdot S_p(r_{d_0}) \cdot \mathrm{PDD}_N(d, r, \mathrm{SSD}) \cdot \mathrm{WF}(d, r_d, x) \cdot \mathrm{TF} \cdot \mathrm{OAR}(d, x) \cdot \left(\frac{\mathrm{SSD}_0 + d_0}{\mathrm{SSD} + d_0}\right)^2}.$$

Dosimetric quantities determination

- Dose per MU under normalization conditions (D'_0)
- Normalized percent depth dose
- Tissue phantom ratios
- Scatter to collimator S_c
- Scatter to phantom S_p
- Tray factor (TF)
- Wedge factor (WF)

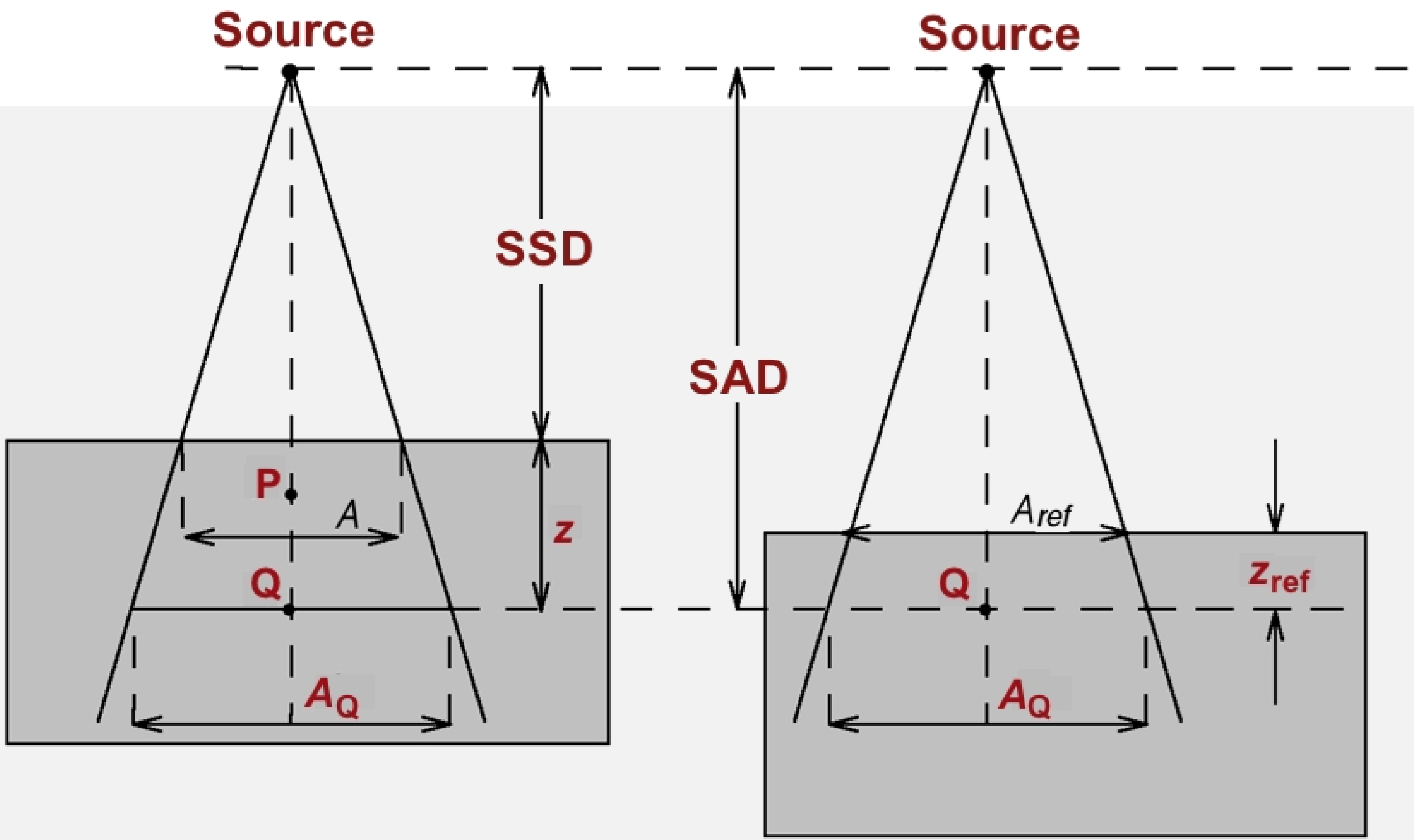
Depth correction: PDD

$$\mathrm{PDD}(z, A, f, h\nu) = 100\frac{D_{\mathrm{Q}}}{D_{\mathrm{P}}} = 100\frac{\dot{D}_{\mathrm{Q}}}{\dot{D}_{\mathrm{P}}}$$



Depth correction: Tissue Phantom Ratio

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



Depth correction

TPR vs PDD

PDD
- are directly measured
- depend by SSD → SSD Setup

TPR
- are unpratical to measure
- are SSD independent → SAD Setup

Depth correction
TPR vs PDD

TPR can computed from PDD table using

$$\mathrm{TPR}(d, r_d) = \left(\frac{\mathrm{PDD}_N(d, r, \mathrm{SSD})}{100\%}\right)\left(\frac{\mathrm{SSD}+d}{\mathrm{SSD}+d_0}\right)^2 \times \left(\frac{S_p(r_{d_0})}{S_p(r_d)}\right).$$

Normalized percent depth dose

- PDD_N, is defined as the percentage ratio of the dose rate at depth to the dose rate at the normalization depth in a water phantom.
- PDD_N are measured using many devices:
  - Cilindrical ionization chamber
  - Plan parallel ionization chamber
  - Diods
  - Microdiamonds

Normalized percent depth dose: effective point of measure

- If a cylindrical or spherical ionization chamber is used, the effective point of measurement of the chamber must be taken into account
- the complete depth ionization curve be shifted to shallower depths (i.e., upstream) by a distance proportional to r_{cav}, where r_{cav} is the radius of the ionization chamber cavity. For photon beams, the shift is taken as $0.6\ r_{cav}$
- No shift in depth-ionization curves is needed if well-guarded plane-parallel ionization chambers are used

Normalized percent depth dose: effective point of measure

The effective point of measurement of cylindrical ionization chambers differs from their geometric center. The exact shift depends on chamber construction details, above all the chamber size, and to some degree on the field-size and beam quality. It generally decreases as the chamber dimensions get smaller

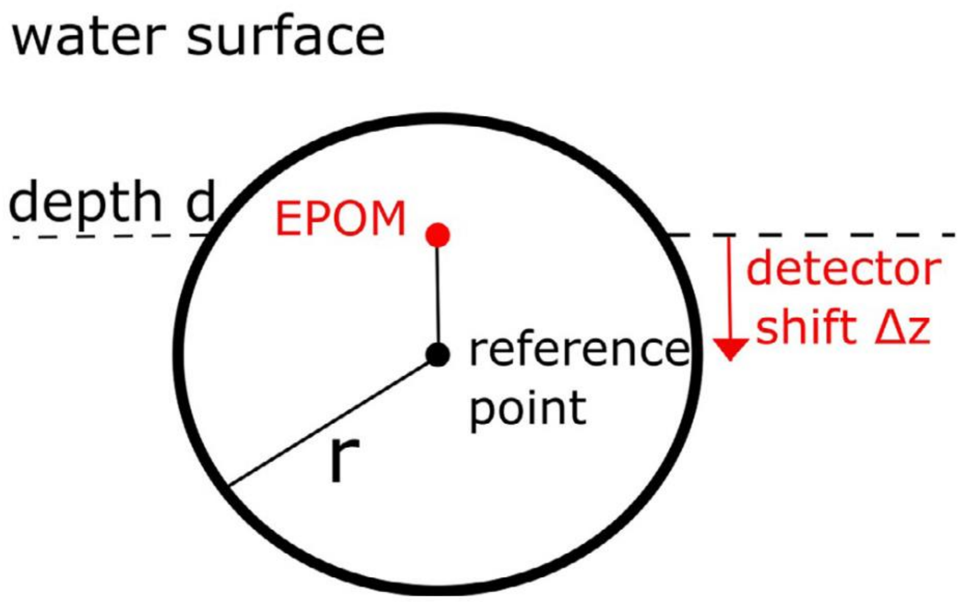


FIG. 1. Schematic illustration how the effective point of measurement (EPOM) is defined. The cylindrical ionization chamber is positioned with its EPOM at a certain depth. This position is reached by shifting the reference point in the detector center by a distance Δz away from the radiation source. [Color figure can be viewed at wileyonlinelibrary.com]

Normalized percent depth dose: effective point of measure

- The perturbation effects of the air cavity can be assumed to a reasonable accuracy to be independent of depth for a given beam quality and field size.
- The depth-ionization curve can thus be treated as depth dose curve for photon beams.
- PPD_N data should be acquired for a series of field sizes ranging from the smallest to the largest field to be used clinically
- The number of measurements should be sufficient such that PDD_N varies by less than 3% between any two measured field sizes

Depth correction
TPR vs PDD

PDD at different SSD can be derived using

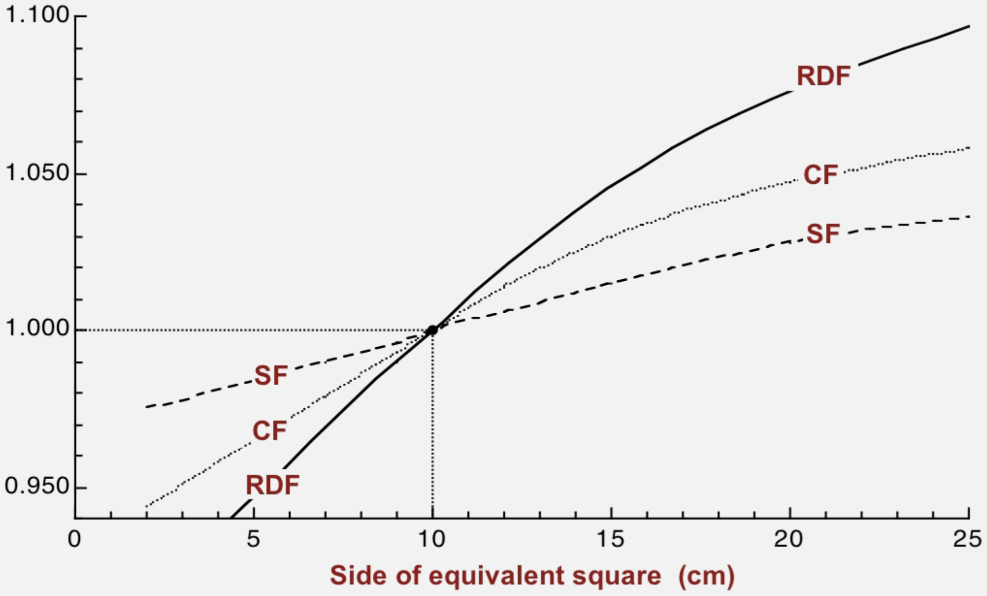
$$\frac{\mathrm{PDD}_N(d, r, \mathrm{SSD}_2)}{\mathrm{PDD}_N(d, r, \mathrm{SSD}_1)} = F \cdot \frac{\mathrm{TPR}(d, r \cdot f_2(d))}{\mathrm{TPR}(d, r \cdot f_1(d))} \cdot \left[\frac{S_p(r \cdot f_1(d_0))}{S_p(r \cdot f_1(d))} \cdot \frac{S_p(r \cdot f_2(d))}{S_p(r \cdot f_2(d_0))} \right]$$

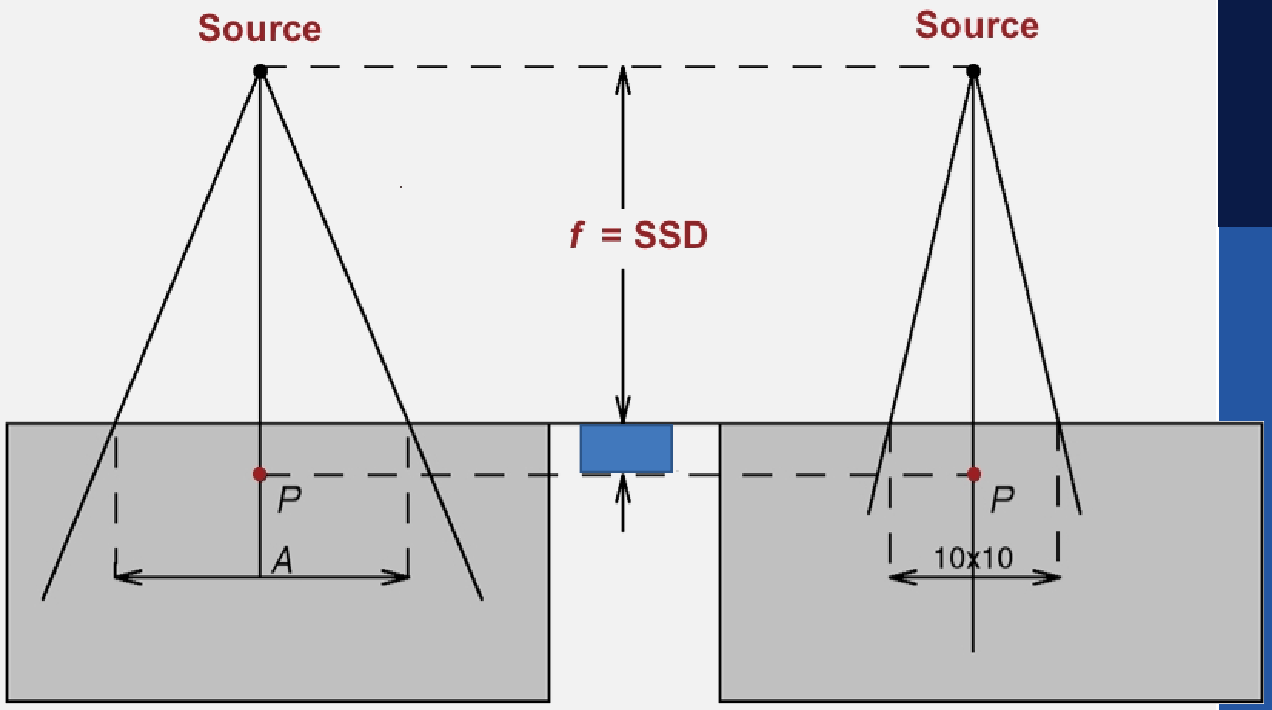
where the Mayneord factor is

$$F = \left(\frac{\mathrm{SSD}_2 + d_0}{\mathrm{SSD}_2 + d} \cdot \frac{\mathrm{SSD}_1 + d}{\mathrm{SSD}_1 + d_0} \right)^2$$

Field size dependence
Output factor

Dose ratio at normalization depth:
A field and normalization field $S_{cp}(A,d_0,f)=S_c*S_p=D(d_0,A)/D(d_0,10x10)$





Scatter to collimator: S_c

- S_c is the ratio of in-air radiation output for a given collimator setting to that for a collimator setting of 10×10 cm^2 at normalization depth
- Measurement set-up for S_c using a mini-phantom

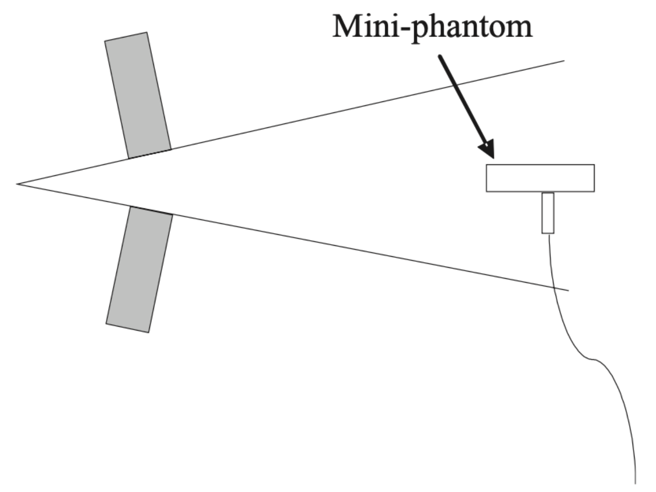


FIG. 5. Diagram illustrating measurement setup for S_c. The cylindrical mini-phantom is aligned coaxially with the central axis of the beam, with the ion chamber positioned at the source-detector distance corresponding to the chosen normalization conditions. The field size is maintained large enough to ensure coverage of the mini-phantom, and other scattering materials are removed from the treatment field.

Scatter to collimator: S_c

- The thickness of material perpendicular to the beam direction should provide enough lateral scatter
- This task group recommends a 4-cm diameter cylindrical mini-phantom coaxial with the central axis of the beam with the detector at 10-cm depth for the measurement of Sc independent of the normalization depth
- Water- equivalent materials are recommended for the construction of the mini-phantom

Scatter to phantom: S_p

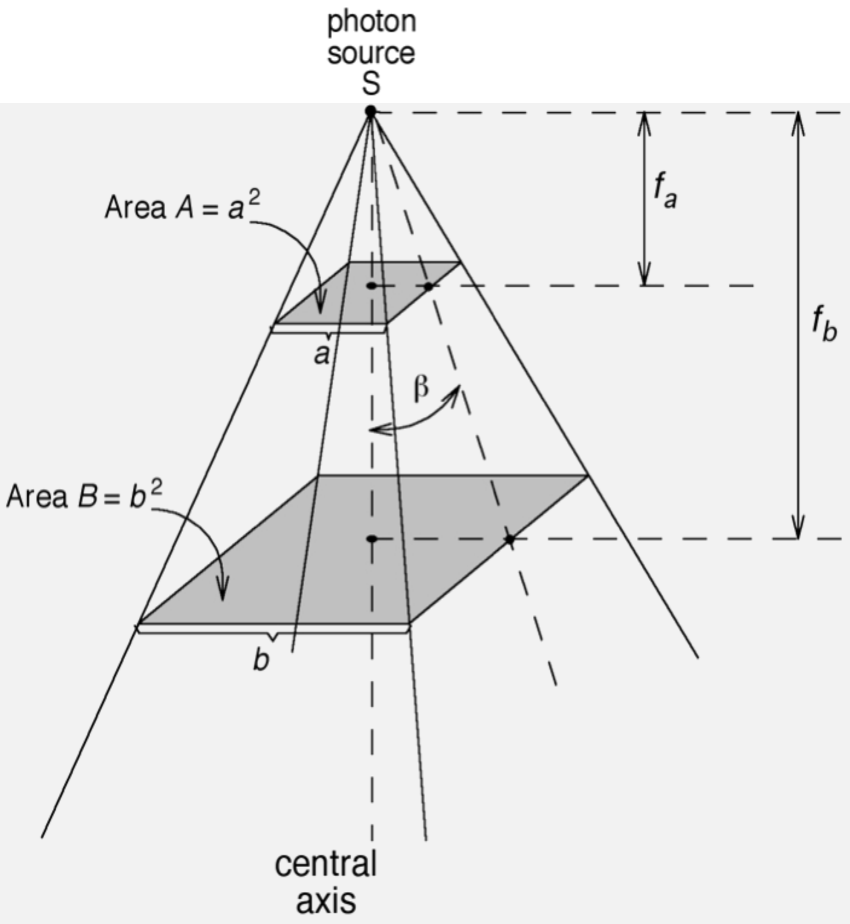
S_p is defined as the ratio of the dose rate at the normalization depth for a given field size in a water phantom to that of the reference field size for the same incident energy fluence. S_p can be computed as a function of the field size at the irradiated volume from the measured quantities S_{cp} and S_c

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

Distance factor
Inverse square law

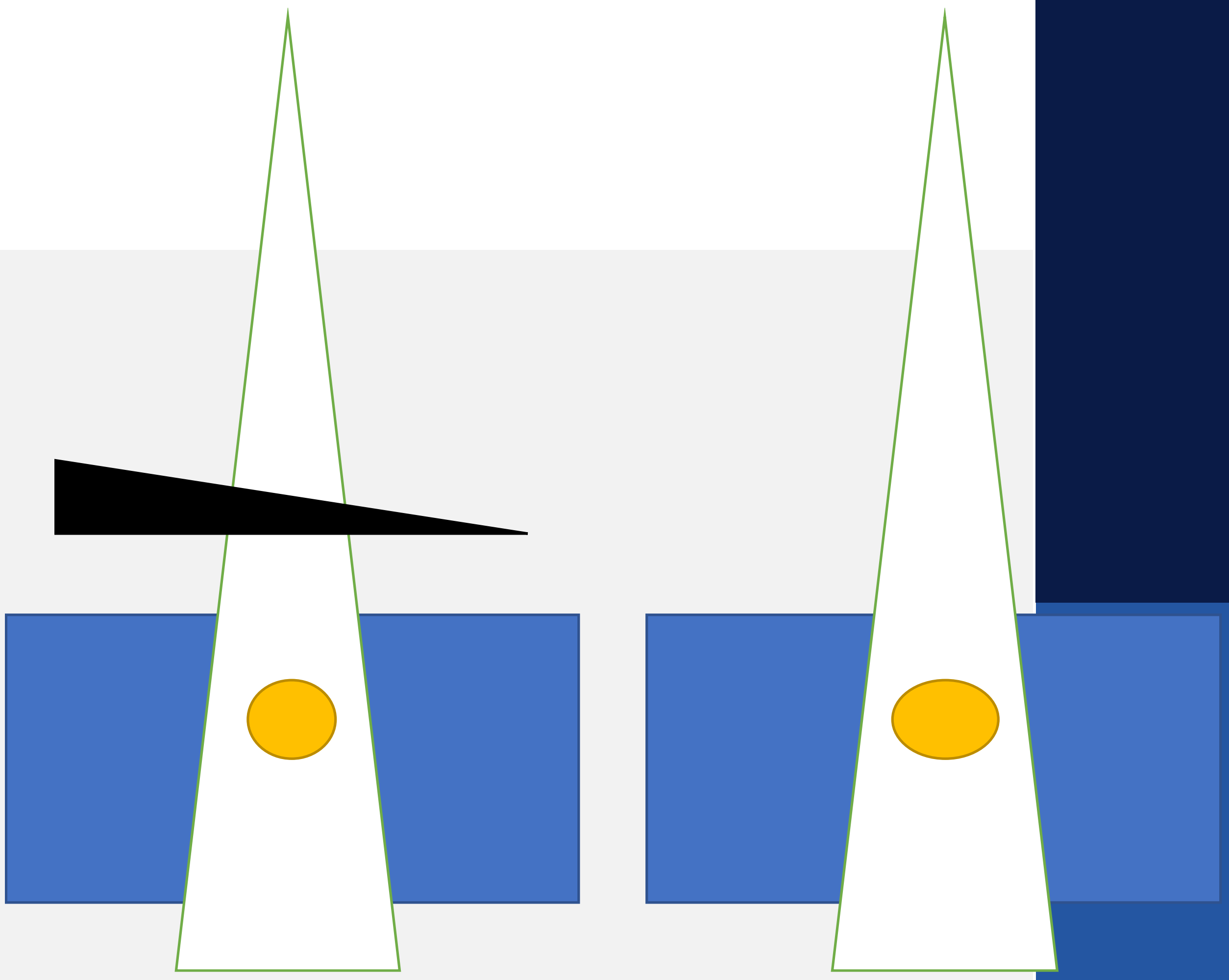
The distance factor is
due to the particle conservation
law

$$\frac{\phi_A}{\phi_B} = \frac{B}{A} = \frac{b^2}{a^2} = \frac{f_b^2}{f_a^2}$$



Wedge factor

- The wedge factor WF is defined as the ratio of the dose rate at the point of calculation for a wedged field to that for the same field without a wedge modifier.
- Physical wedge factors should be measured as a function of both field size and depth
- With the chamber axis perpendicular to the gradient direction of the wedge, two sets of measurements should be made with the wedge in opposite orientations to accommodate uncertainties in the chamber position and wedge mounting.



Field size determination S_c

- S_c is the scatter function due to collimator system
- The effective field size for the S_c depends on

Equivalent square field given by:

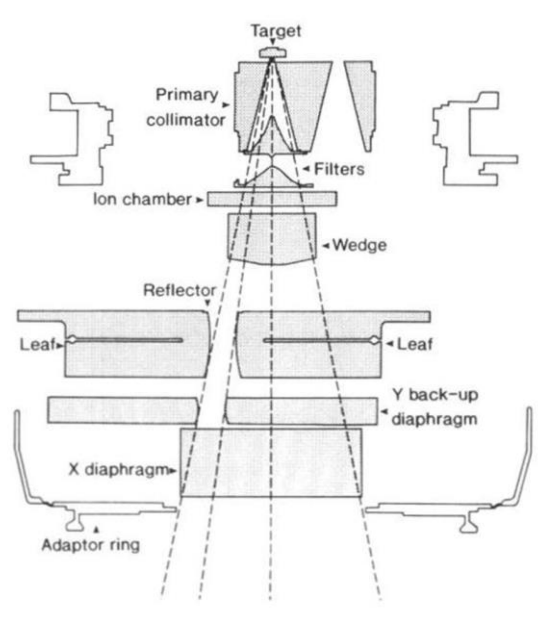
4*Area/Perimeter

$$r_c = 4\left(\frac{r_{jU} \cdot r_{jL}}{2r_{jU} + 2r_{jL}}\right)$$

Collimator exchange effect

The collimators position for shaping x and y beam have a different distance from the X-rays source: this could produce a difference in the S_c by exchanging the the upper lower collimator size:

$$S_c(x,y) \neq S_c(y,x)$$



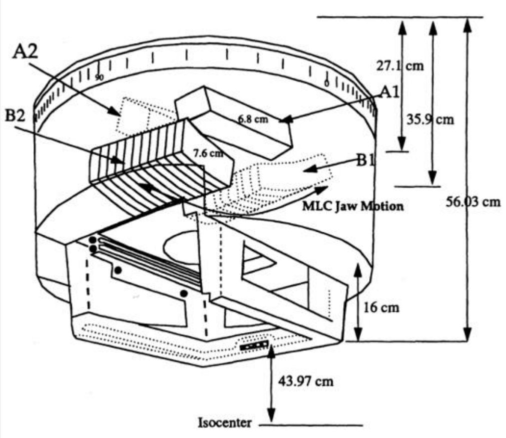


FIG. 2. Schematic diagram of the Siemens MLC head. In this design, the double-focused bank of 54 leaves is mounted in place of the lower collimator. Each of the tungsten leaves is 7.6-cm thick and projects to a 1.0-cm wide radiation field at isocenter. All leaves can be independently moved to an over-travel of 10 cm past the central axis (Ref. 24).

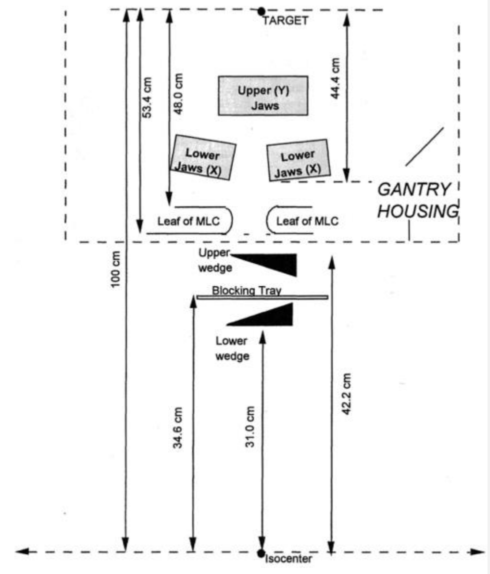
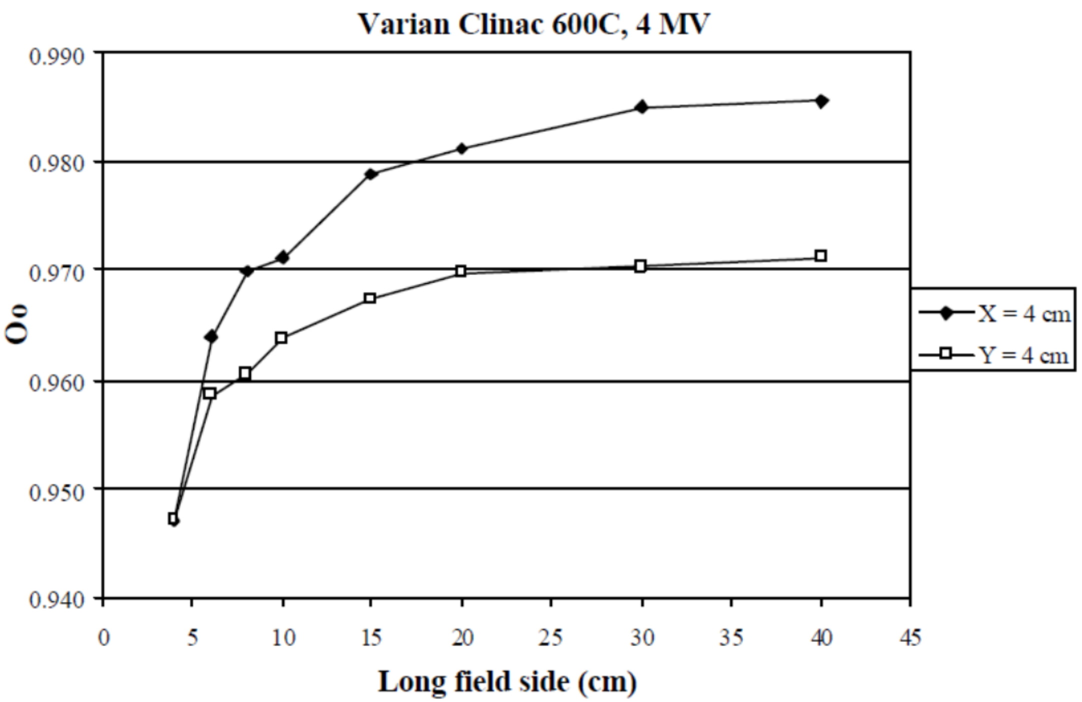
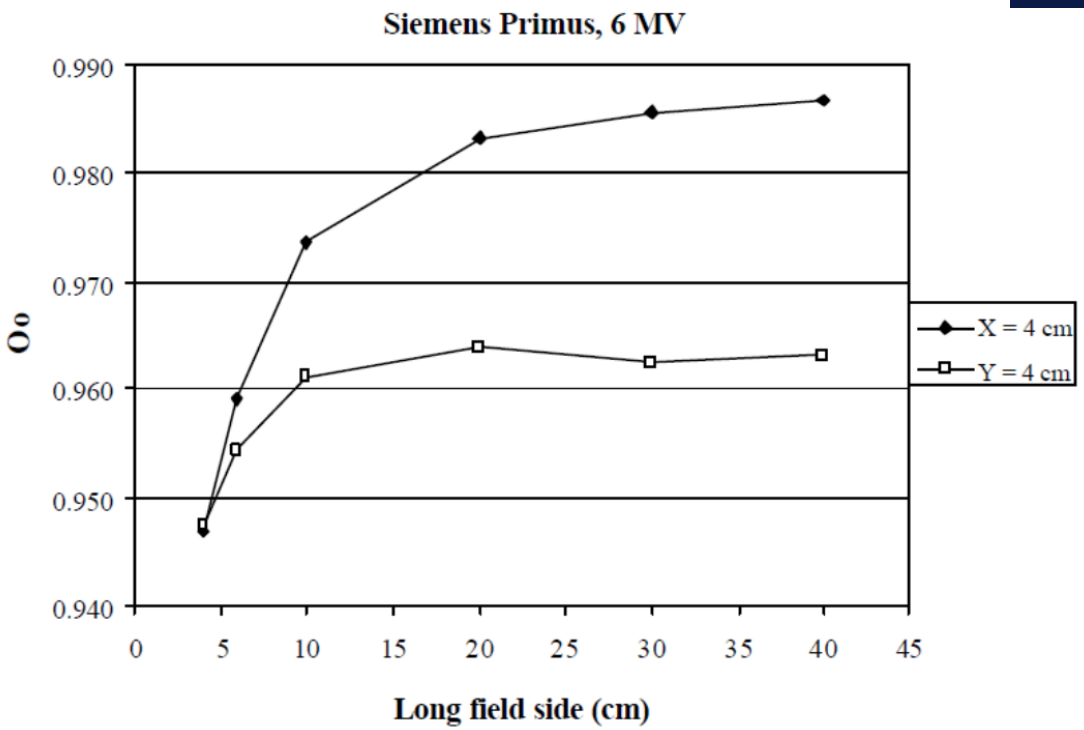


FIG. 3. Cross-sectional view of Varian MLC head for a 2100C accelerator (Ref. 26). In this design, the leaf banks are mounted in carriages placed below the lower collimator, with leaf widths of 0.5- or 1.0-cm projected at SAD, depending on MLC model.

Collimator exchange effect





Collimator equivalent square field

- The collimator equivalent square field C_e takes into account the collimator exchange effect (CEE), i.e. for rectangular fields the output ratios for a given collimator setting are different if the upper and lower collimator jaws are interchanged.
- The magnitude of the CEE, therefore, depends on the construction (flattening filter, collimators, additional shielding...) of the head of the treatment machine (tipically < 2%).

Field size determination

S_p TPR approach

The field-size argument r_d of S_p is the equivalent square of the field size incident on the patient, projected to the depth of the point of calculation.

Thus, unlike Sc, the argument for Sp will change with a change of source-point distance (SPD).

Field size determination

S_p PDD approach

The field-size argument r_{d0} of S_p is the equivalent square of the field size incident on the patient, projected to the normalization depth.

Field size determination
TPR and PDD

- The field size argument for TPR is the equivalent square of the field size incident on the patient, projected to the depth of the point of calculation.
- The field-size argument for PDD_N is the equivalent square of the field size incident on the patient.

Dose per MU under normalization conditions (D'_0)

- The normalization conditions are not necessarily equal to the reference conditions under which the linear accelerator is calibrated
- The normalization point is at depth d_0=10 cm SAD=100 SSD= 90 cm
- It could be still possible to calibrate the dose using an SSD protocol but it you start up a new machine it is better to calibrate at the normalization point
- Normalization with D'_0= 1cGy/Mu is the more pratical