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MU Calculation: the ESTRO formalism

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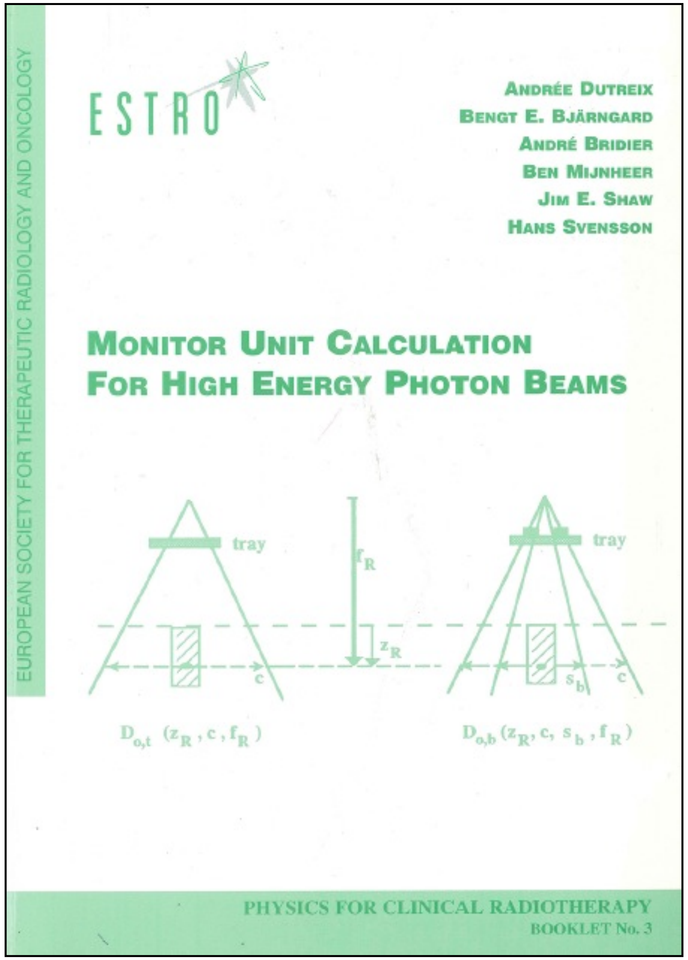
The ESTRO formalism

- In ESTRO Booklet 3 [Dutreix et al 1997] a formalism has been developed to calculate MU's for radiation treatments with photon beams provided by accelerators and ^{60}Co units.

- The IAEA was also involved in the work; the first draft was outlined by a consultants' group in Vienna in 1992. Responsible IAEA officer was Hans Svensson (Sweden).

- The formalism is applicable to most practical situations met in radiotherapy applying rectangular, blocked and wedged beams, both under isocentric and fixed source-skin distance conditions.



The Rationale

- The basis for the procedure is the determination of the absorbed dose per MU under <u>reference conditions</u>:
  - 10 cm depth in water, source-detector distance equal to

    a) the isocentre distance (generally 100 cm) and a 10cm x 10cm field size at this distance,
- or

    b) the regular source-skin distance (generally 100 cm) and a field size of 10cm x 10cm at this distance.
- The traditional MU calculation using dosimetric quantities referring to <u>dose maximum</u> has been replaced by a formalism which applies quantities referring to measurements <u>at 10 cm depth</u> for all photon beam qualities.
- The reason for this change is that the maximum dose depends on the degree of electron contamination that varies critically with change in beam geometry.

Beam Modelling (Estro booklet 10)

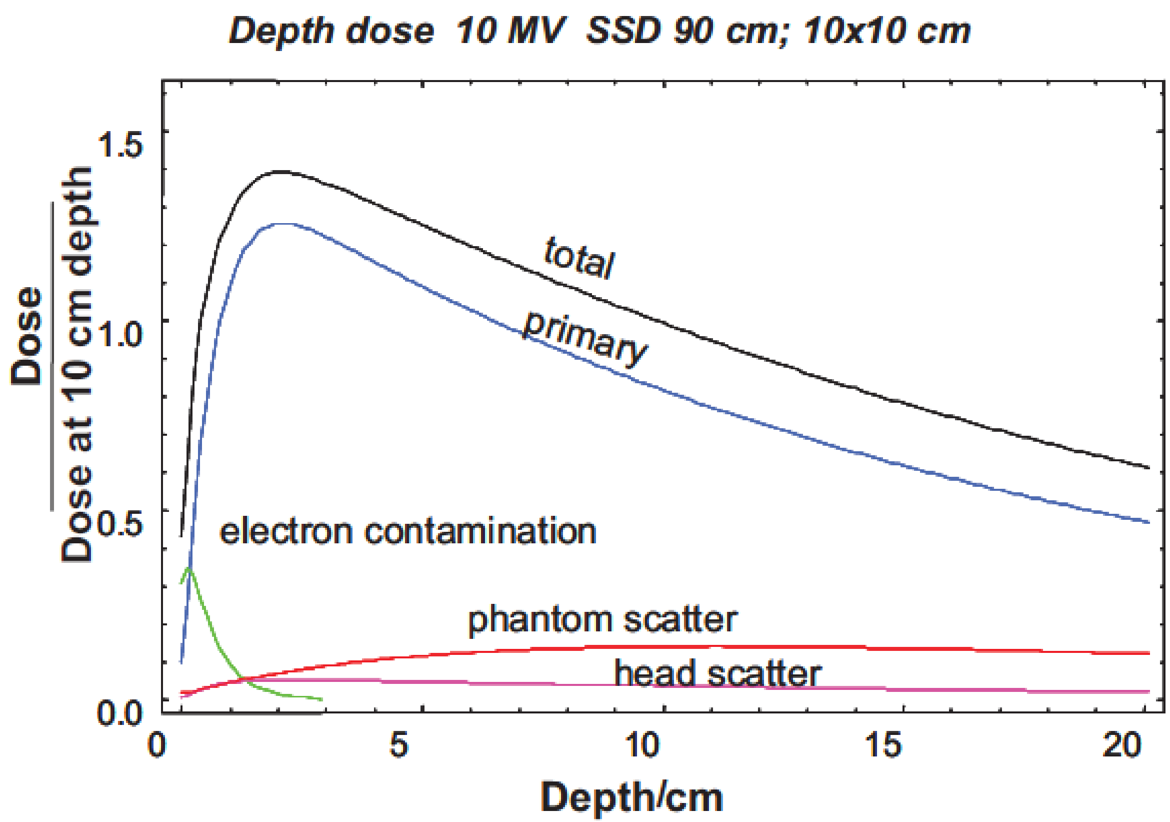


Figure 5.9 Depth dose distributions for a 10×10cm² field of 10 MV photons, showing separately the direct beam primary dose (blue), direct beam phantom scatter dose (red), electron contamination (green), total head scatter dose (pink) and the total sum of all components (black). Normalization is versus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

Effect of electron contamination

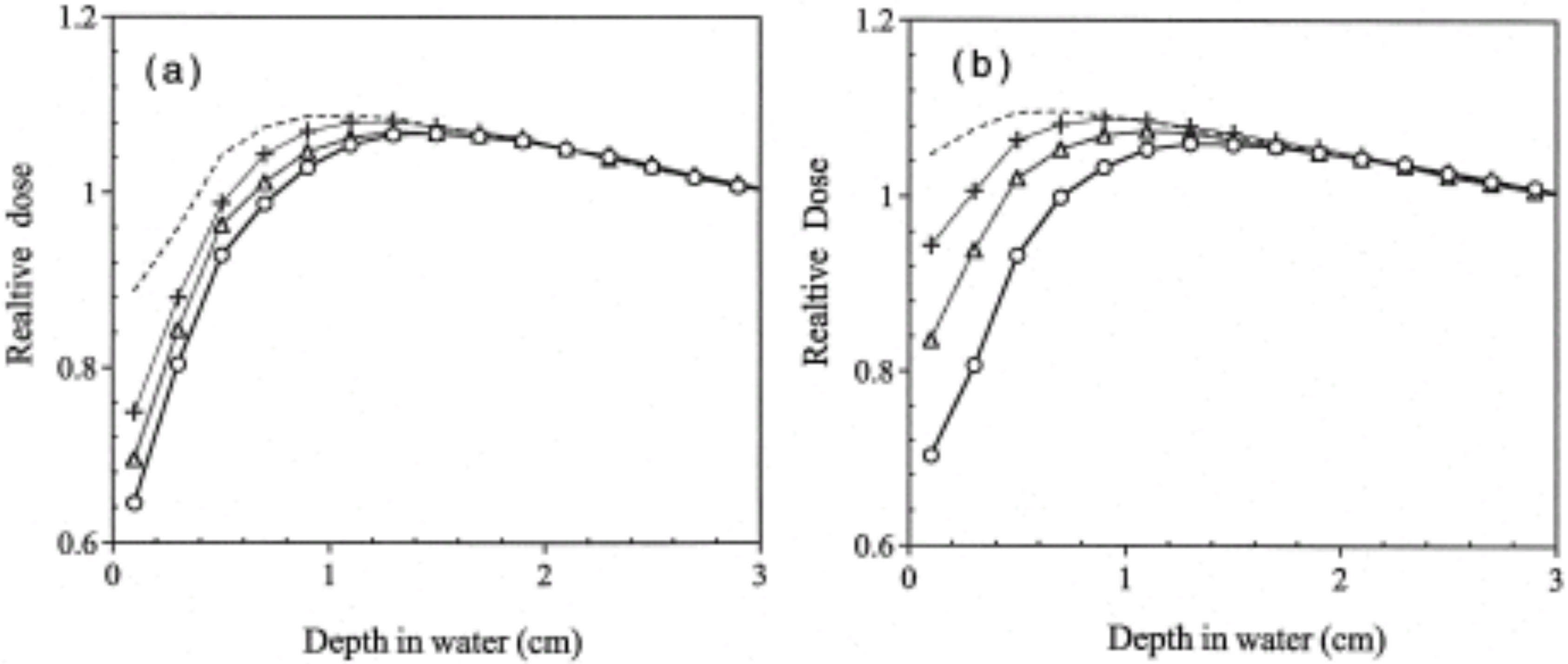


Fig. 2. Dependence of measured central axis depth dose on the spoiler-to-surface distances for two square fields (a) 5 × 5 cm² and (b) 10 × 10 cm². Open beam (circles) and STSD of 6 cm (dashed line), 10 cm (plus signs), and 15 cm (triangles).

Calculation Methods

- The use of data measured in a mini-phantom for several irradiation geometries in addition to large water phantom measurements is recommended.
- It is possible in this way to separate the contribution to the dose due to scatter in the accelerator (or ^{60}Co-unit) head and due to scatter in the water phantom [e.g. van Gasteren , 1991].

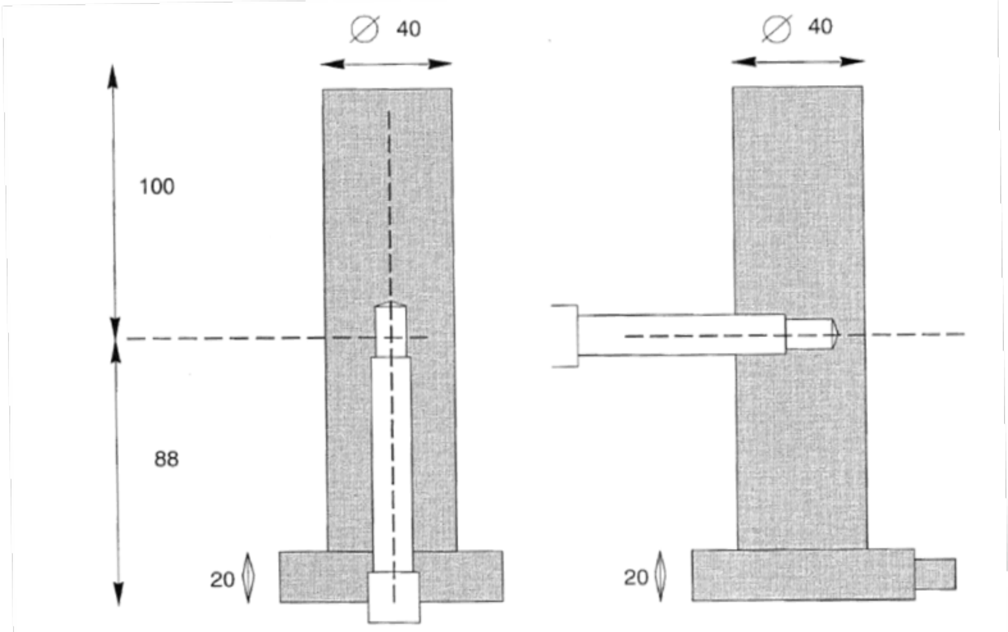


Figure 8-1. Examples of plastic mini-phantoms; left) a small ionization chamber is positioned with its axis parallel to the beam axis; right) a small ionization chamber is inserted perpendicular to the beam axis;

Mini-phantom Measures

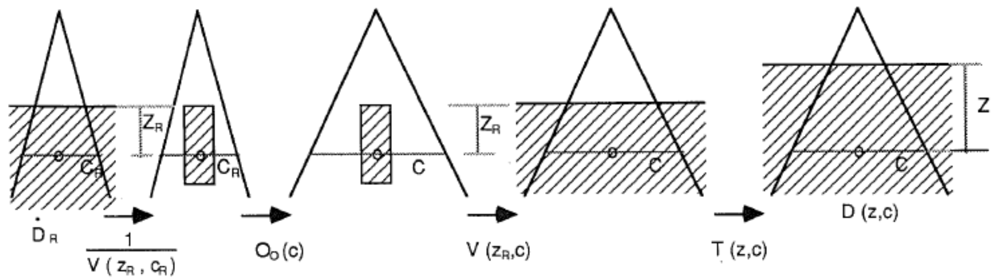
- Water phantoms are generally used to measure the absolute dose to the point. This involves the primary and secondary components.
- This type of water phantom has the disadvantage in measuring the head scatter factor.
- When increasing the field size for measurement, the irradiated volume also increases in larger field sizes. So, the phantom scatter contribution will be high.
- In the mini-phantom, the irradiated volume is same for all field sizes.
- So it measures only the primary and collimator scatter of the secondary components and prevents the scatter contribution from the phantom.
- Mini phantom must have adequate wall thickness to stop the contaminating electrons from passing through the sides of the mini phantom to the chamber.

Calculation Methods

- The starting point of the ESTRO formalism is a beam calibration at the reference point.
- Then, measurement data obtained in the reference geometry, are used either in isocentric or fixed source-skin distance conditions.
- Thus, 2 sets of equations are derived and their mutual relationship is described.

FROM:

ISO SETUP



TO:

Figure 5-4 : Conditions of measurement for the derivation of the dose **D(z,c)**, in isocentric conditions, from the dose per monitor unit or per unit of time $\dot{\mathbf{D}}_\mathbf{R}$ in reference conditions, and the output ratio $\mathbf{O_0(c)}$ in a mini-phantom.

SSD SETUP

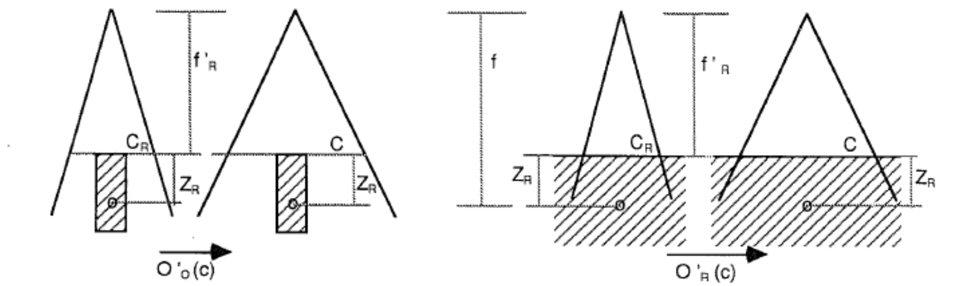


Figure 5-5 : Conditions of measurement of the output ratios $\mathbf{O'_0(c)}$ and $\mathbf{O'_R(c)}$, at a fixed source-skin distance, as the ratios of the dose for a collimator setting **c** and a collimator setting $\mathbf{c_R}$, in a mini-phantom and in full scatter conditions, respectively.

The ESTRO Booklets

- ESTRO Booklet 3 provides the formalism, the definition of the physical quantities as well as the equations for MU calculation.
- These equations take into account all possible physical effects influencing the dose delivery at a specific point.

- ESTRO Booklet 6 provides numerical data required for applying the equations for monitor unit calculation.
- Data are provided for a ^{60}Co-unit and 4, 6, 10 and 18 MV beams of 4 different types of accelerator.
- Recommendations are given for the measurements required to apply the formalism.
- Finally a number of examples are given.

Lesson topics

- In this lesson we will present the equations that are required to illustrate the application of the formalism in clinical practice.
- We will restrict ourselves to isocentric conditions, the most commonly applied treatment set-up, thus limiting the number of formulae.
- Equations are now required to determine the dose

$$\mathbf{D(z,c)}$$

- under treatment conditions, at depth z, for field size c, for open, wedged, and blocked fields.
- Starting point will be the dose per MU along the central beam axis under reference conditions,

$$\mathbf{D_R}$$

- determined in a large water phantom.

Equations: Open beams

$$D(z,s_e) = \dot{D}_R \bullet U \bullet O_0(c_e) \bullet \frac{V(z_R,s_e)}{V(z_R,c_R)} \bullet T(z,s_e),$$

- D_R: dose per MU under reference conditions
- U: number of monitor units

- where the output ratio $O_0(c)$ accounts for variations in <u>head scatter</u>, and the last two terms for attenuation and scattering variations in the large water phantom.

- In open beams this separation of the different physical components is not essential, but <u>it facilitates the dose calculation in more complex situations when shielding blocks are used</u>.

Equations: Open beams

$$D(z,s_e) = \dot{D}_R \bullet U \bullet O_0(c_e) \bullet \frac{V(z_R,s_e)}{V(z_R,c_R)} \bullet T(z,s_e),$$

- $O_0(c_e)$: output ratio determined in a mini-phantom for field size c_e (equivalent to the head scatter factor S_c)
- c_e: collimator equivalent square for a rectangular collimator setting (X,Y)
- c_R: reference field size defined by the collimator (10 cm x 10 cm field size at isocentre)
- z_R: reference depth (10 cm is recommended)
- $\frac{V(z_R,s_e)}{V(z_R,c_R)}$: the ratio of volume scatter ratios at the reference depth z_R for field sizes s_e and c_R (equivalent to the phantom scatter factor S_p)

Equations: Open beams

$$D(z,s_e) = \dot{D}_R \bullet U \bullet O_0(c_e) \bullet \frac{V(z_R,s_e)}{V(z_R,c_R)} \bullet T(z,s_e),$$

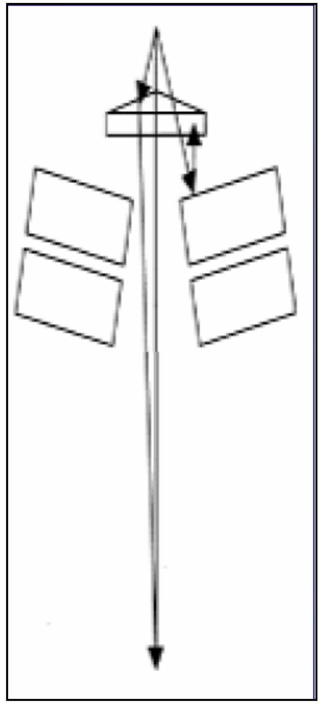
- $T(z,s_e)$ tissue-phantom ratio at depth z for field size s_e for use with phantom scatter
- s_e : equivalent square for use with phantom scatter

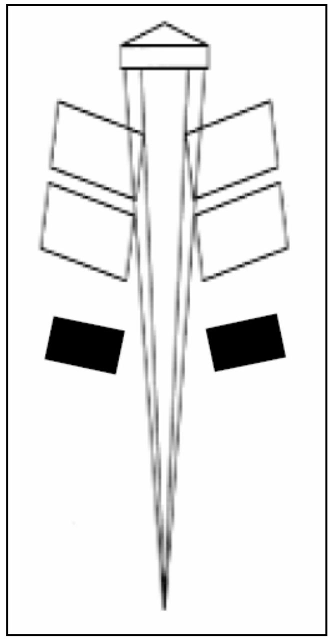
$$s_e = 2 \bullet X \bullet Y / (X + Y)$$

Sterling equation

Collimator Exchange Effect

- 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 effect originates from differences in energy fluence of photons originating from the flattening filter reaching the point of interest and from different amounts of radiation scattered backwards from the upper and lower collimator jaws into the beam monitor chamber.
- The magnitude of the CEE, therefore, depends on the construction (flattening filter, collimators, additional shielding…) of the head of the treatment machine (tipically < 2%).





JP Gibbons

Collimator equivalent square

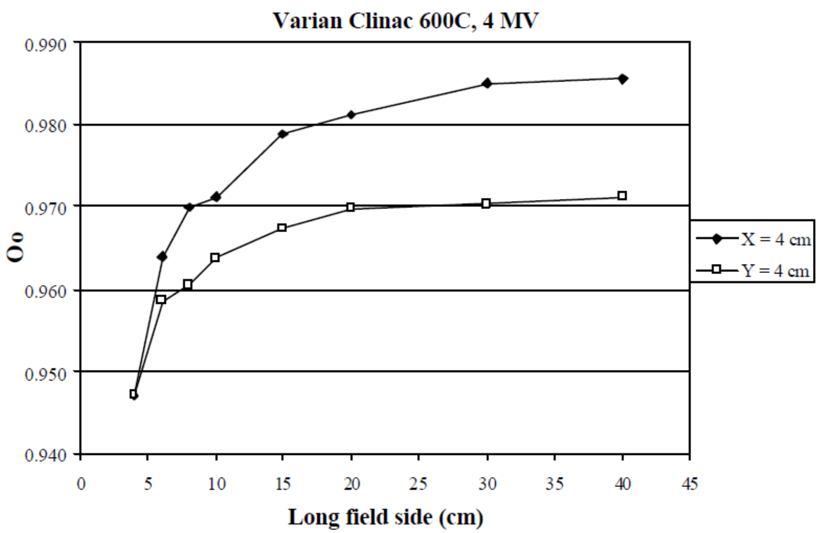
- If a separation of the output factor is applied in a collimator scatter part, i.e. the output ratio O_0 determined in a mini-phantom, and in a phantom scatter part, i.e. the ratio of volume scatter ratios $V(z_R,s_e)/V(z_R,c_R)$, then the CEE can be fully attributed to O_0.

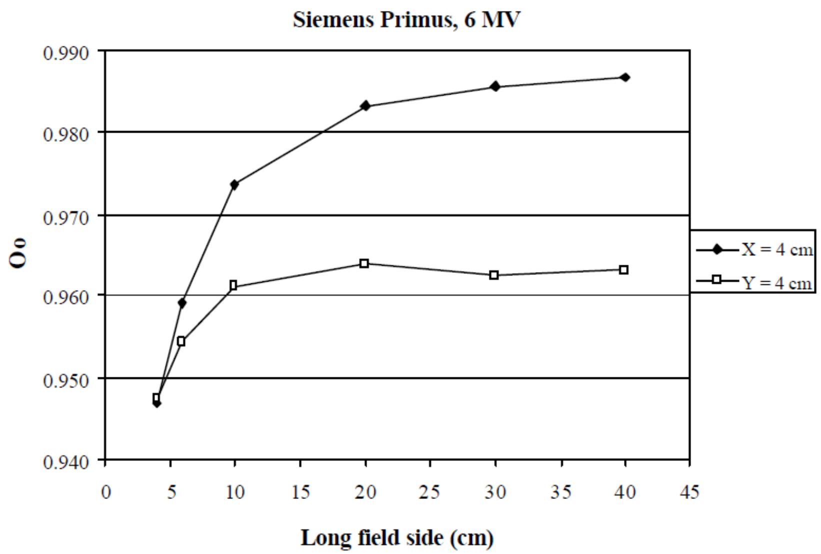
- For a rectangular field setting (X,Y), where X and Y are the openings of the lower and upper jaws respectively, c_e can be derived by using an equation proposed by Vadash and Bjärngard [1993]:

$$c_e = (A + 1) \bullet X \bullet Y / (A \bullet X + Y),$$

- where **A** is the relative weight of the X- and Y- collimator settings, specific for each treatment unit and beam quality.

- **A** may be different for the open and wedged beams of the same nominal energy



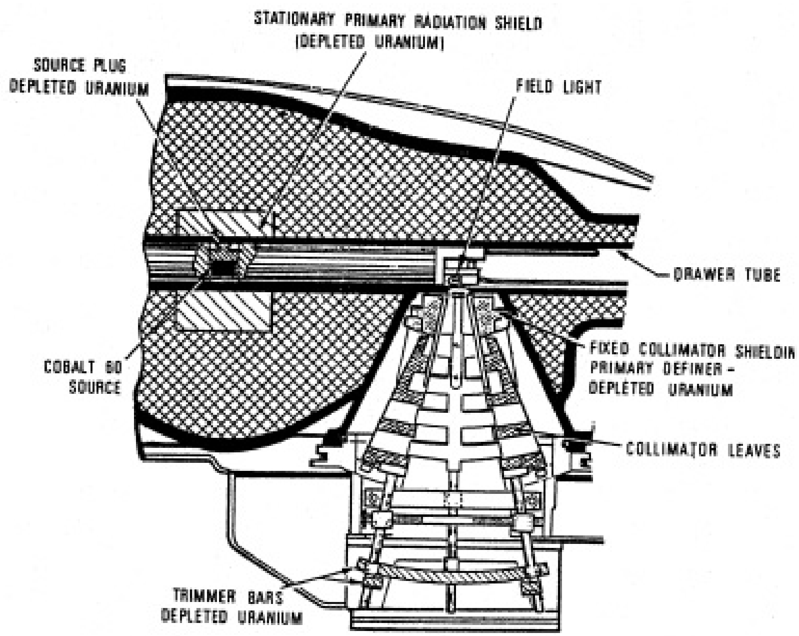


O_0 is plotted as a function of the long field side, keeping either the X- or the Y- collimator fixed at 4 cm.

Table 4.4 A factors for the four x-ray beams.

Machine	*Nominal accelerator potential*	*A factor measured*	*A factor calculated*	
			Yu et al Eq. (4.2)	*Kim et al Eq. (4.3)*
Varian Clinac 600	4 MV	1.7	1.7	1.5
Siemens Primus	6 MV	1.7	1.8	1.5
GE-CGR Saturne 41	10 MV	1.4	2.0	1.7
EOS SL20	18 MV	1.4	1.65	1.5

CEE for Co-60 Units

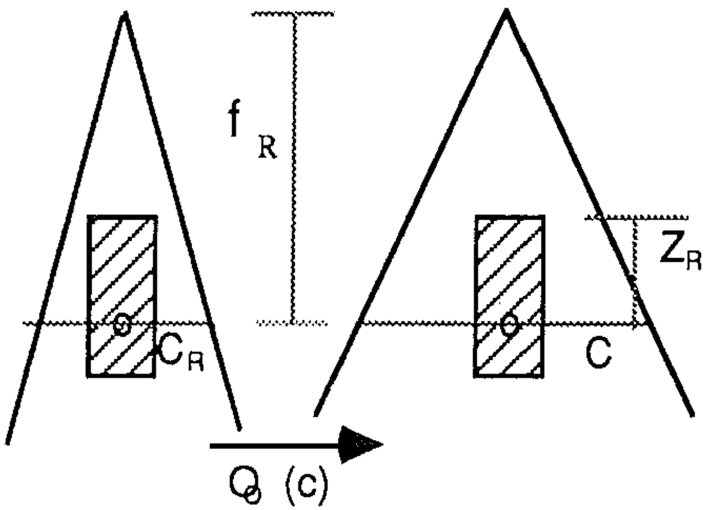


MDS Nordion Theratron-780, ESTRO Booklet 6

- The difference in distance between the source and the X- and Y- collimator parts of the head of a ^{60}Co treatment unit is relatively small.
- Consequently, the CEE is ***negligible*** in clinical practice (generally within 0.5%).
- Implicitly, the phantom scatter part remains unchanged if the X- and Y- collimator setting is interchanged for the setup of a rectangular field.

OUTPUT RATIO O_0

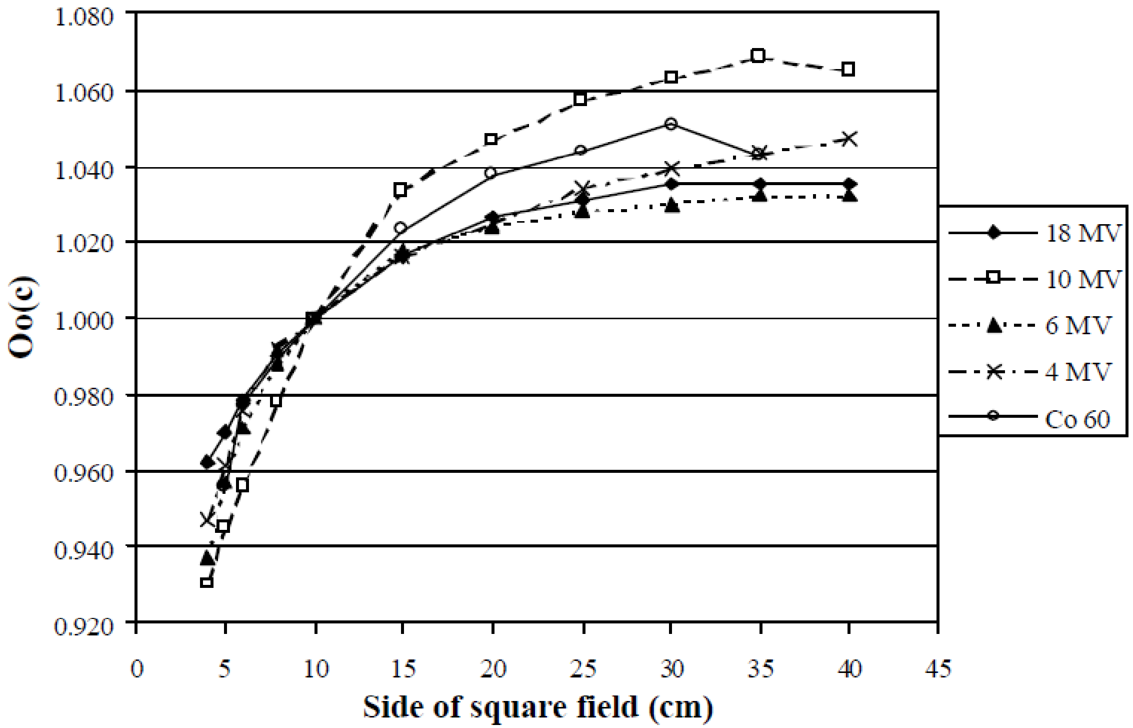
- It is defined as the ratio of the absorbed dose at the reference depth z_R for filed size c, to the dose at the same depth for the reference field size c_R, measured in a mini-phantom, where both c and c_R are defined at the reference distance f_R.



$$O_R(c) = \frac{D(z_R, c)}{D(z_R, c_R)}$$

- The output ratio O_0 can be considered to be equivalent to the Khan's *head scatter factor S_c* ; <u>however</u>, O_0 values are measured at 10 cm water equivalent depth in a mini-phantom, while Khan defined the head scatter factor at the depth of dose maximum.

OUTPUT RATIO O_0



S Senthilkumar and Ramakrishnan, JMP 2008

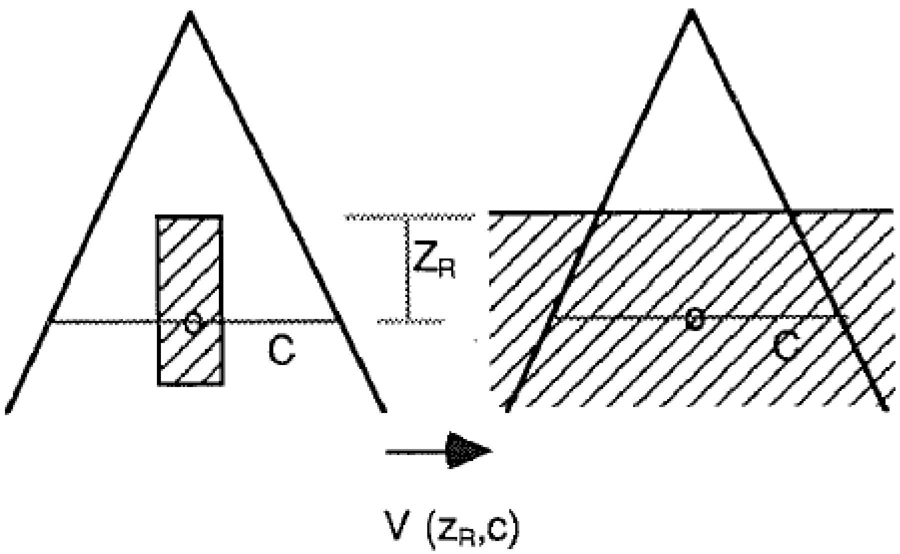
O_0 variation with field size strongly depends on the treatment head design.

The maximum variation is observed for the GE-CGR Saturne 41 beam, where the flattening filter is much wider and is positioned at a more downstream position compared with other machines.

PHANTOM SCATTER CORRECTION FACTORS

- To describe the contribution to the dose of the phantom scattered photons, a new quantity is introduced, the **Volume scatter ratio V**, conceptually similar to the tissue-air ratio, but the dose in air is now a quantity which can be easily measured.

- Volume scatter ratios, **V**, are the ratios of the dose values measured under full scatter condition and in a mini-phantom.



$$V(z_R,c) = D(z_R,c)/D_0(z_R,c)$$

PHANTOM SCATTER CORRECTION FACTORS

- $V(z_R,s)$ expresses the influence of the phantom scatter on the dose at a specific calculation point.
- <u>It depends on the field size **s** at the depth of measurement</u>, but is not, in a 1st approximation, a function of the source-detector distance, provided that the 2 doses are measured at the same distance.

$$V(z,s)=V(z,c)$$

- only when the distance to the source is the reference distance f_R !
- The ratio of $V(z_R,s)$ and $V(z_R,c_R)$ represents the contribution of the phantom scatter at the reference depth z_R when the beam size varies from c_R to c.

$$\frac{V(z_R,s_e)}{V(z_R,c_R)}$$

PHANTOM SCATTER CORRECTION FACTORS

- The ratio of $V(z_R,s)$ and $V(z_R,c_R)$ is equal to the *phantom scatter correction factor*

$$\frac{V(z_R,s_e)}{V(z_R,c_R)} = S_p(z_R,s)$$

- as defined by several groups, e.g. Khan and van Gasteren et al.

- This ratio is > or < 1, depending on whether c is > or < c_R.

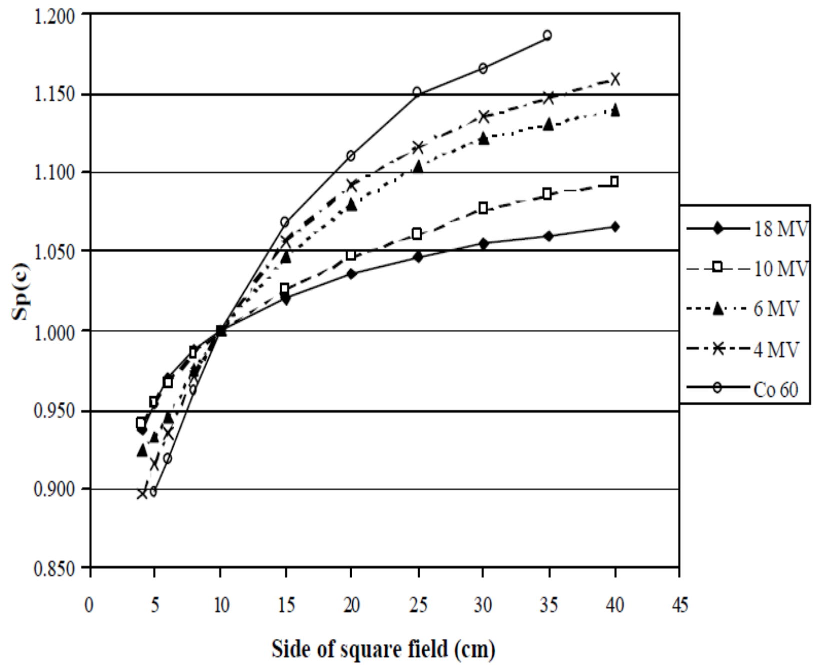
- It depends on the <u>beam quality</u> $T_{20/10}$, but is not very sensitive on the type of accelerator or to the radial energy variations.

- Based on experimental data obtained from a large number of linear accelerators, a complete set of S_p factors was constructed by Storchi and van Gasteren <u>as a function of field size and quality index</u>.

PHANTOM SCATTER CORRECTION FACTORS

- The original S_p data of Storchi and van Gasteren were defined for the fixed SSD set-up, i.e. with field size definition at the phantom surface at 100 cm from the x-ray source.

- These data have been adapted to calculate phantom scatter correction factors for field sizes used in the isocentric formalism.



- The data presented by Storchi and van Gasteren showed that, within the experimental uncertainty, which is less than ~ 1%, the S_p curves of different machines with the same quality index coincide.

- The ESTRO Group recommends to perform always S_p measurements for small field sizes for each beam of a treatment unit.

TISSUE-PHANTOM RATIOS

- T(z,s) is the ratio of the dose D(z,s,f) at the depth z and the dose D (z_R, s, f) at the depth z_R for the same field size s at f and same source-point of interest distance f:

$$T(z,s) = \frac{D(z,s,f)}{D(z_R,s,f)}$$

- Under clinical conditions, T(z,s) does not depend on **f**, but is a function of the field size **s** at **f**.
- In ESTRO Booklet 6 tissue-phantom ratio data are given for the five photon beams under consideration.
- These data were obtained from percentage depth dose (PDD) data provided by each institution according to the conversion described by Dutreix et al. (next slide).

TISSUE-PHANTOM RATIOS

- Measured PDD values $P(z,s,f_R)$ have been renormalized to <u>reference depth</u> dose data,

$$P_R(z,s,f_R) = P(z,s,f_R) \,/\, P(z_R,s,f_R)$$

- which were converted to T values according to:

$$T(z,s) = P_R\,(z,s',f_R) \bullet \frac{V(z_R,s')}{V(z_R,s'')} \bullet \frac{(f_R+z)^2}{(f_R+z_R)^2}$$

where

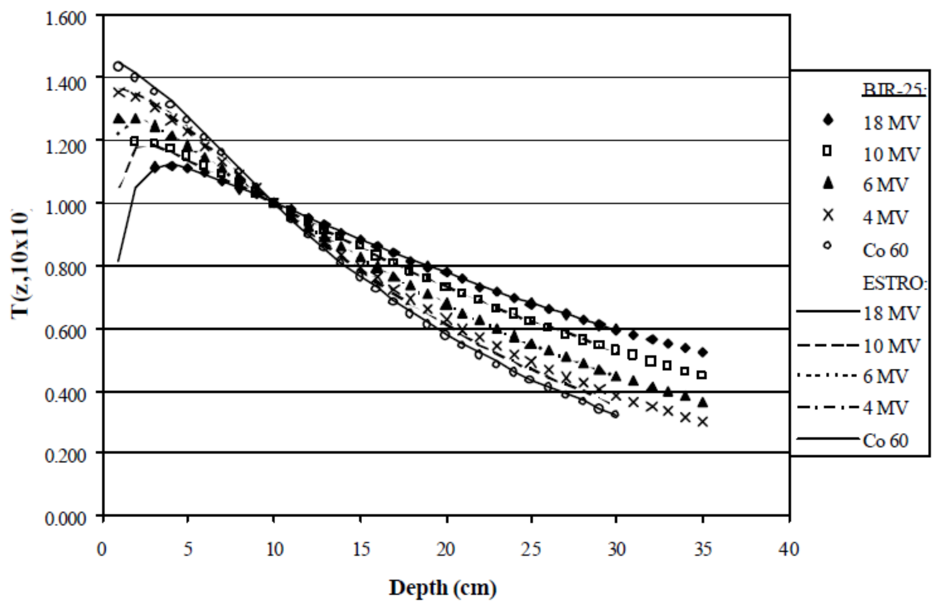
s is the field size at the isocentre and equal to the collimator setting c

s' is equal to $s \bullet (f_R - z) / f_R$

s'' is equal to $s \bullet (f_R - z_R) / f_R$

Tissue-phantom ratios

- Brit. J. Radiol. Suppl. 25 provides data of tissue-maximum ratios.
- By taking the ratio of two tissue-maximum ratios, tissue-phantom ratios can be obtained.
- T values obtained in this way have been compared with the ESTRO data.
- In general these data sets agree with each other within 2%.



- An exception is the 4 MV x-ray beam, where deviations from -1% at shallow depths up to +6% at a depth of 30 cm are observed. These deviations are given as “local dose values”.

WEDGED BEAMS

- The dose under treatment conditions D(z,c,w) of a wedged beam can be derived from the dose per MU under reference conditions D_R , the output ratio and the tissue phantom ratio of the open beam, by introducing a field size dependent wedge factor k_w.
- This leads to the following equation:

$$\boxed{D(z,c,w) = \dot{D}_R \cdot U \cdot O_0(c) \cdot \frac{V(z_R,c)}{V(z_R,c_R)} \cdot k_w(z,c) \cdot T(z,c).}$$

- $k_w(z,c)$ wedge factor determined in a large water phantom.

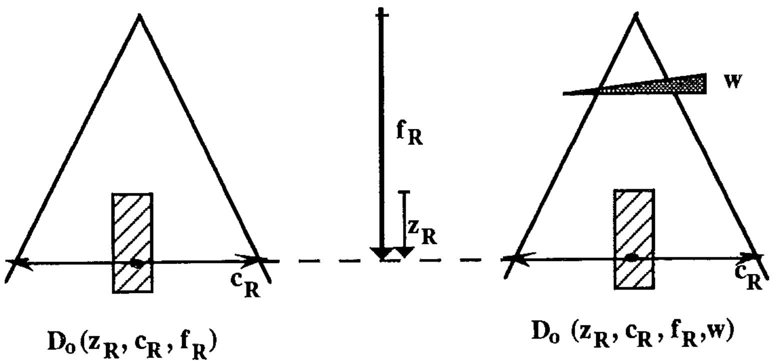
  It is a function of field size and depth.

WEDGED BEAMS

- A more common approach is to define a wedge factor at the reference depth z_R and to take its depth dependence into account by the tissue-phantom ratio T(z,c,w) at depth z for field size c with the wedge in the beam, yielding the following equation:

$$D(z,c,w) = \dot{D}_R \cdot U \cdot O_0(c,w) \cdot \frac{V(z_R,c,w)}{V(z_R,c_R)} \cdot k_{o,w}(c_R) \cdot T(z,c,w),$$

- where $k_{o,w}(c_R)$, the wedge factor determined in a <u>mini-phantom</u> under reference conditions, takes into account the modifications of the head scatter produced across the wedge filter.



$$k_{o,w}(c_R) = \frac{D_0(z_R, c_R, f_R, w)}{D_0(z_R, c_R, f_R)}$$

Wedge Factors

- The relation between the wedge factor determined in a large water phantom or a mini-phantom is given by:

$$k_w(z_R,c_R) = k_{o,w}(c_R) \bullet \frac{V(z_R,c_R,w)}{V(z_R,c_R)}.$$

- For most situations with high energy photon beams,

$$V(z_R,c) = V(z_R,c,w),$$

- i.e., insertion of the <u>wedge will not modify the phantom scatter</u> correction factor considerably. Consequently, in those cases:

$$k_{o,w}(c_R) = k_w(z_R,c_R)$$

- At <u>low energy beams</u>, differences can be found and care has to be taken with the assumption of equivalence of $V(z_R,c)$ and $V(z_R,c,w)$ [Georg 1999].

Wedged Beams

- For the high energy beams, the equation can now be rewritten as:

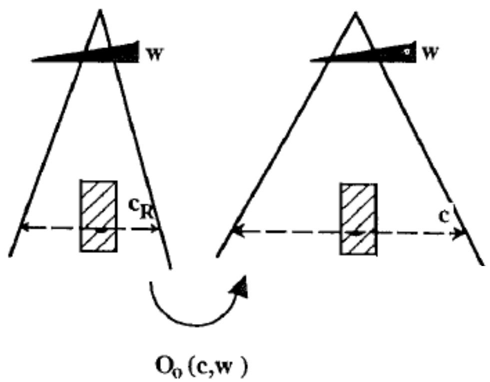
$$D(z,c,w) = \dot{D}_R \cdot U \cdot O_0(c,w) \cdot S_p(c) \cdot k_w(z_R,c_R) \cdot T(z,c,w),$$

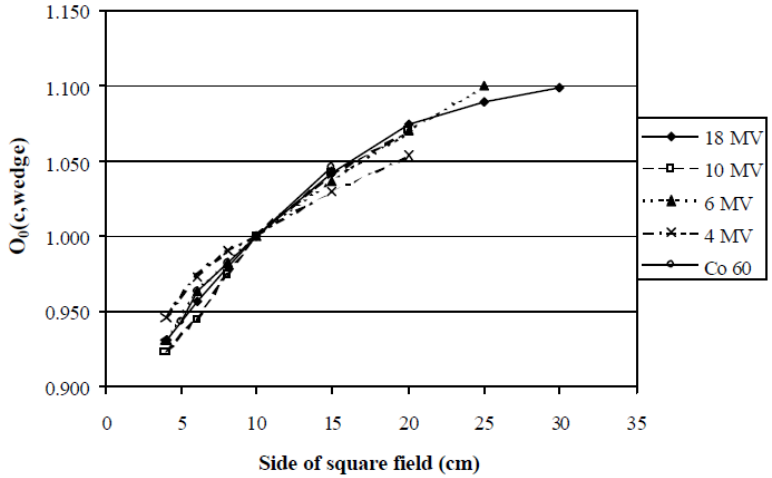
and

$$D(z,c,w) = \dot{D}_R \cdot U \cdot O_0(c,w) \cdot S_p(c) \cdot k_{o,w}(c_R) \cdot T(z,c,w),$$

- in which the ratio of the volume scatter ratios is equal to the phantom scatter correction factor <u>S_p(c) for the open beam</u>.

Wedged Output ratio





- O_0(c,w) values for five beam qualities with a 45° or 60° (in case of internal wedge) wedge in the beam are given in ESTRO Booklet 6, as a function of the side of the square field.

- Larger variations of output ratios with field sizes are observed with a wedge than without a wedge.

- An overall variation of the O_0 values of the order of 17% is observed for the wedged 18 MV beam when the side of the square field is varied from 4 cm to 30 cm. The corresponding variation for the open field is 7%.

Wedged T(z,c,w)

- The variation of the wedge factor with depth is due to beam hardening or softening.
- This variation is taken into account by the change in tissue-phantom ratio of the wedged beam compared with the open beam.
- In ESTRO Booklet 6, for the 10 and 18 MV beams the difference between the TPR of the wedged and the open beam is almost negligible in practice,
- but for the lower beam qualities the ratio of the wedge and open beam TPR is continuously increasing with depth and varies significantly.

Wedged T(z,c,w)

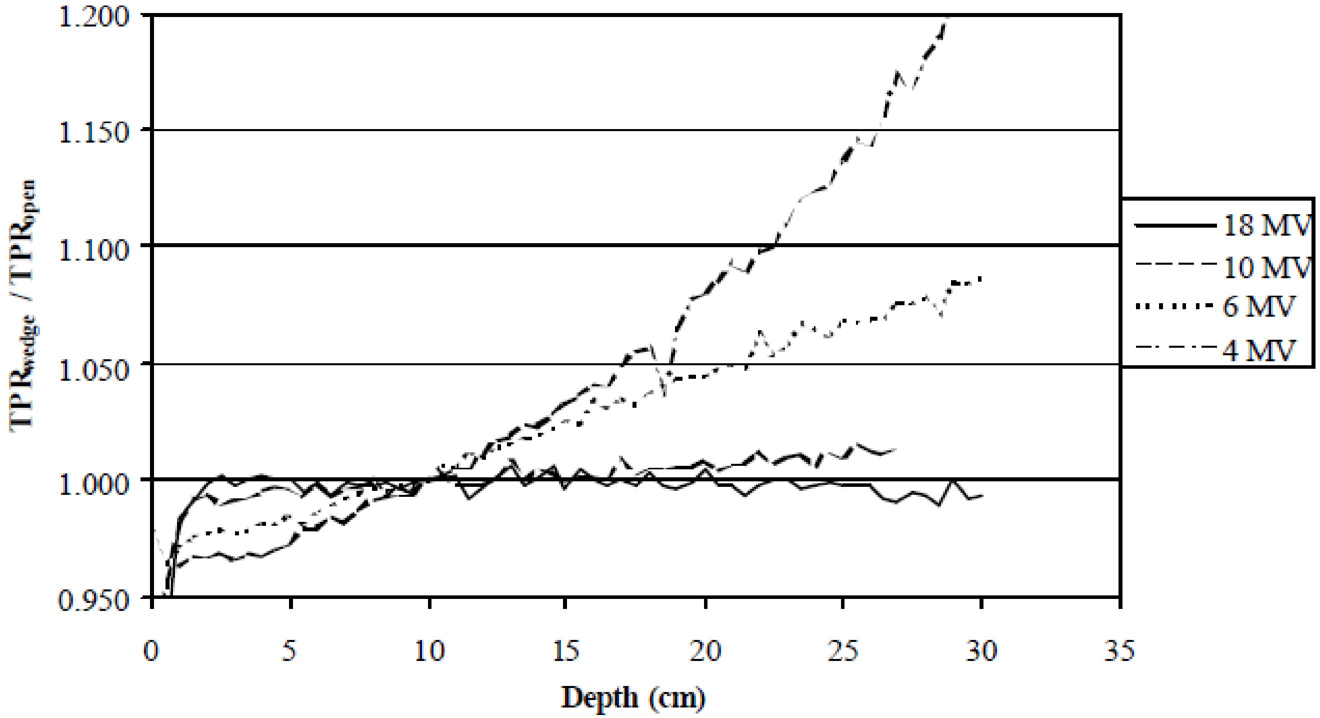


Figure 5.4 Ratio of tissue-phantom ratios of wedged and open beams, determined for a 10 cm x 10 cm field. The variation in the curves illustrates the uncertainty in the measured data. The data are measured with the wedges listed in Table 5.1.

BLOCKED BEAMS

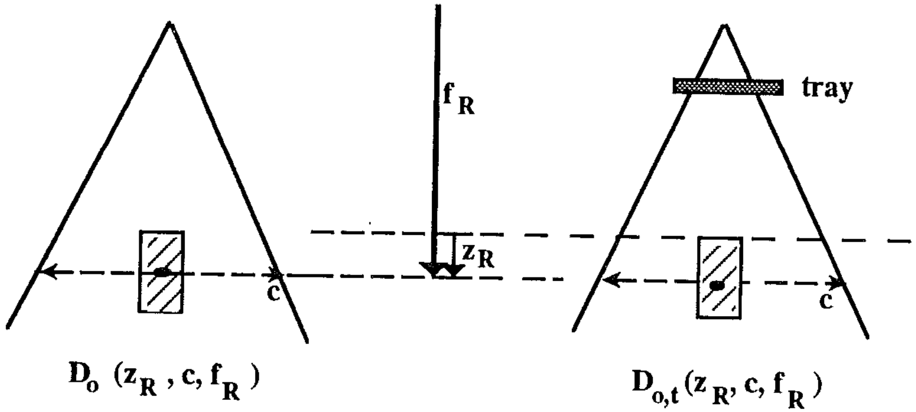
- The dose under treatment conditions D(z,c,s b) of a blocked beam can be derived from the dose per MU under reference conditions D_R according to the following equation:

$$D(z,c,s_b) = \dot{D}_R \bullet U \bullet O_0(c) \bullet \frac{V(z_R,s_b)}{V(z_R,c_R)} \bullet k_{o,t}(c) \bullet k_{o,b}(c,s_b) \bullet T(z,s_b,b)$$

- s_b field size defined by the shielding blocks at the point of interest.
- $k_{o,t}(c)$ tray factor measured with a mini-phantom
- $k_{o,b}(c,s_b)$ correction for the presence of the shielding blocks

Tray Factor $k_{o,t}(c)$

- It is defined as the ratio of the dose measured in the mini-phantom for field size c under reference conditions, with and without the shadow tray, for the same number of monitor units.



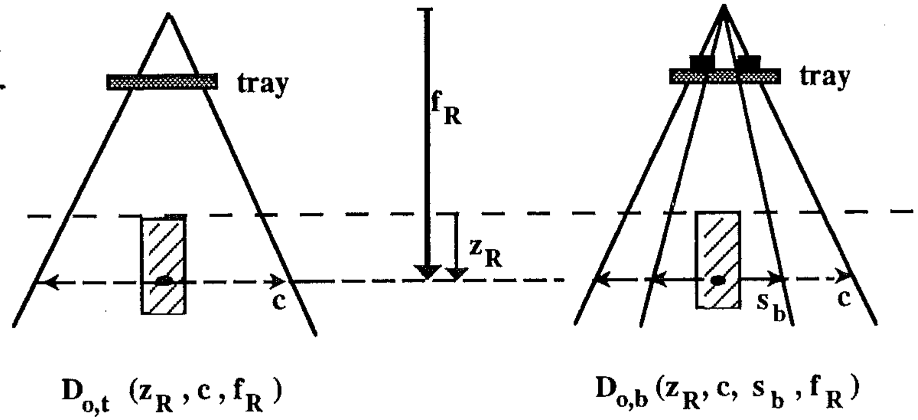
- It is assumed independent of the distance to the source. It depends on the photon beam quality, on the depth of measurement and slightly on the collimator opening because of the additional photons scattered by the tray.
- Note that $\mathbf{k_{o,t}(c) \cong k_t(c)}$

the tray transmission factor measured in a large water phantom

Block Correction Factor

- $k_{o,b}(c,s_b)$ correction for the presence of the shielding blocks in the beam determined with the mini-phantom. **c** is the collimator defined field size and $\mathbf{s_b}$ is the field size defined by the shielding blocks, both at the isocentre.

$$k_{o,b}(c,s_b) = \frac{D_{o,b}(z_R, c, s_b, f_R)}{D_{o,t}(z_R, c, f_R)}$$

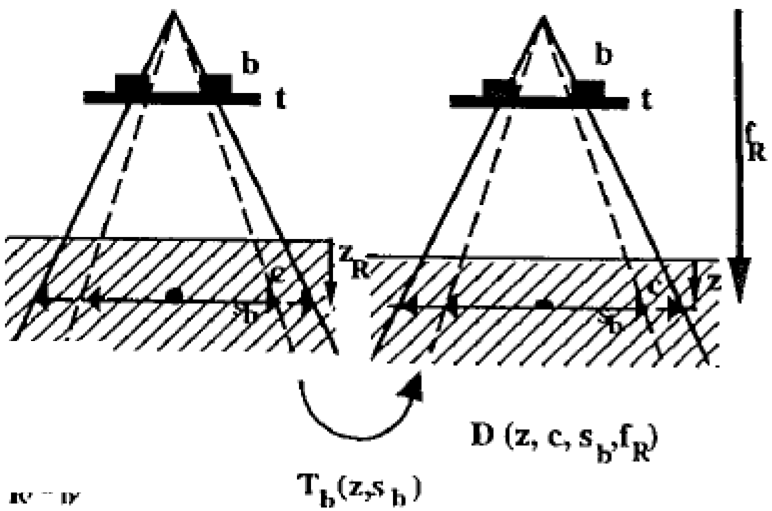


- It represents the variation of head scatter, when placing shielding blocks on the tray.
- For most photon beam energies, $k_{0,b}$ does not differ from 1 by more than 2%.

Blocked TPR

- $T(z, \mathbf{s_b}, b)$: tissue-phantom ratio measured with the shadow tray in the beam

- It takes into account the effect of the tray and the shielding blocks on the depth dose.
- <u>It can be approximated</u> by the tissue-phantom ratio measured in the <u>open beam</u> for the same field size $\mathbf{s_b}$ for $z > z_R$.
- For $z < z_R$, T values can be modified by the presence of a tray and should be checked for the tray-to-skin distances in practical use.



NON-ISOCENTRIC TREATMENTS

- For treatments performed at a distance **f** which is different from the reference distance $\mathbf{f_R}$ but for otherwise identical treatment conditions (same depth and field size at the point of interest), only a modification in the primary photon fluence has to be taken into account.

- The following equation is presented for the case of an open photon beam, based on the <u>application of the inverse square law to the dose in the mini-phantom under reference conditions</u>:

$$D(z,s_e,f) = \dot{D}_R \cdot U \cdot O_0(c_e) \cdot \frac{V(z_R,s_e)}{V(z_R,c_R)} \cdot T(z,s_e) \cdot \left(\frac{f_R}{f}\right)^2$$

which can be rewritten as:

$$D(z,s_e,f) = D(z,s_e,f_R) \cdot \frac{O_0(s_e \cdot f_R/f)}{O_0(s_e)} \cdot \left(\frac{f_R}{f}\right)^2$$

- $\mathbf{s_e}$ treatment field size at distance f
- $\mathbf{c_e}$ collimator field size at f_R, which is equal to $s_e \bullet f_R / f$

SUMMARY OF THE MEASUREMENT OF THE BASIC BEAM DATA

Quantity	*Field description*	*Square fields*[1]	*Rectangular fields*	*Source-detector distance*	*Phantom*[2]
P,T	open	+	+[3]	100	fsp
	wedged	+	+[3]	100	fsp
	tray	+	-	100 and 80	mp or fsp
O_R	open	+	+[3]	100	fsp
	wedged	+	+[3]	100	fsp
O_0	open	+	+[3,4]	100	mp
	wedged	+	+[3]	100	mp
	tray	+	-	100	mp
$k_{o,w}(c_R)$ or $k_w(c_R)$	wedged/open	+	-	100	mp or fsp
$k_{o,t}(c)$ or $k_t(c)$	tray/open	+	-	100	mp or fsp
$k_{o,b}(c,s_b)$	tray/blocks	+	+	100	mp

fsp = full scatter phantom; mp = mini-phantom

It is recommended to perform additional measurements in a number of test situations to check and verify the methodology of MU calculation.

Measure of Tissue-Phantom Ratios T

- In practice depth dose curves (PDD) are more easy to measure than T values.
- Consequently, a conversion from the measured PDD values to T values can be applied.
- Measured PDD values $P(z,s,f_R)$ can be renormalized to reference depth dose data,

$$P_R(z,s,f_R) = P(z,s,f_R) / P(z_R,s,f_R)$$

- which are then converted to T values according to:

$$T(z,s) = P_R(z,s',f_R) \bullet \frac{V(z_R,s')}{V(z_R,s'')} \bullet \frac{(f_R+z)^2}{(f_R+z_R)^2}$$

where

s	field size at the isocentre and equal to the collimator setting c
s’	$= s$ Ÿ $(f_R - z) / f_R$
s”	$= s$ Ÿ $(f_R - z_R) / f_R$

Mini-phantoms

- The diameter of the rod phantom should be as small as possible to avoid side scatter.
- It is recommended to use a diameter ≥ 4 cm for most photon beams in clinical use.
- Build-up caps of high-Z materials (brass, iron etc) cause a larger scatter for fieldsizes > 30 cm.
- It is recommended to use the polystyrene, PMMA or water-filled mini-phantom for measurements «in air».

Discussion and Conclusion

- The measurement-based ESTRO formalism is applicable to most practical situations encountered in RT applying rectangular, blocked and wedged beams, both under isocentric and fixed source-skin distance conditions.
- The accuracy of the ESTRO formalism is stated to be around 1-2% for the supported beam geometries, making it attractive as a basis for independent dose calculations.
- At the present time, however, the formalism does not include asymmetric fields, off-axis calculations, dynamic wedges and entrance dose calculations, though several papers are available with appropriate integrations.
- Despite these shortcomings, the formalism proposed by ESTRO has the potential to become the unifying method with which to aid communication between various centres.

(S. Johnsson, 2003)

http://estro-education.org/publications/Pages/ESTROPhysicsBooklets.aspx

Thank you
for your attention !