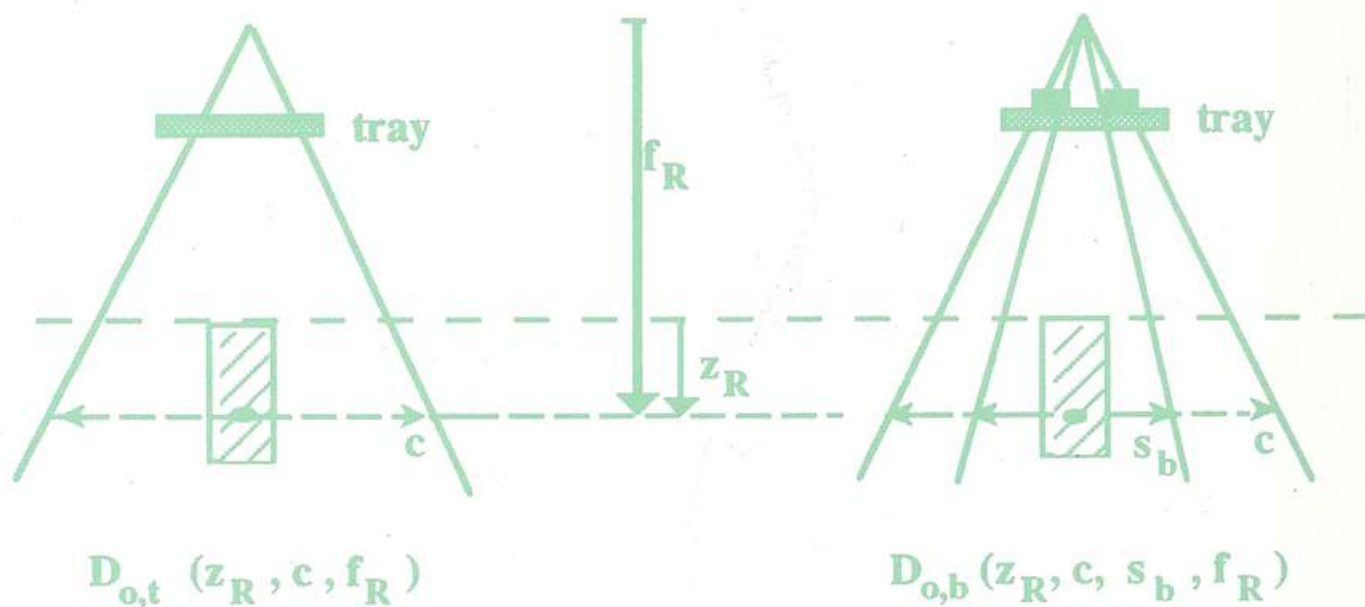




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MONITOR UNIT CALCULATION FOR HIGH ENERGY PHOTON BEAMS



**MONITOR UNIT CALCULATION
FOR HIGH ENERGY PHOTON BEAMS**

PHYSICS FOR CLINICAL RADIOTHERAPY

Booklet No 3.

Aims and scope

Recognizing the importance of accurate dose determination in radiation therapy the International Atomic Energy Agency initiated a programme in dosimetry of ionizing radiation soon after its formation. In 1968 IAEA, in co-operation with WHO, initiated a regular mail dosimetry service, covering 200 to 400 hospitals annually. Secondary Standards Laboratories (SSDLs) were set up and a Network of SSDLs was established in 1976.

The IAEA was also involved in developing and recommending dosimetry procedures in radiotherapy (e.g. Technical Report Series No. 110 and 277). The present paper suggests procedures for dose monitor determinations to be used when photon beams from ^{60}Co -units or accelerators are used in radiation treatments.

The present work was initiated by the IAEA and the first draft was outlined by a consultants' group in Vienna in December 1992. The members of this group were B. Bjärngård (USA), A. Bridier (France), A. Dutreix (Belgium), B. Mijnheer (The Netherlands), J. Shaw (UK), and H. Svensson (responsible IAEA officer). Further consultants' meetings were held in November 1993 (Paris), May 1994 (Belgium) and May 1995 (Vienna).

The core part of this report (chapters 5 to 7) is based on papers by A. Bridier and A. Dutreix (Bridier and Dutreix, 1986; Dutreix et al. 1994). The final ideas have been developed during these consultants' meetings and also in meetings sponsored by the European Society for Therapeutic Radiology and Oncology (ESTRO). A workshop with 110 participants, representing much of the international expertise in this field, was organized by ESTRO (directors : A. Dutreix and H. Svensson) in October 1995 (Italy) to discuss the draft report. Suggestions given at this meeting were introduced in the final draft.

Monitor unit calculation for high energy photon beams

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Foreword.

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Summary and conclusions.

A procedure for monitor unit calculation for radiation treatments with photon beams from accelerators and ^{60}Co -units is recommended. The formalism can be used for most practical situations met in radiotherapy, e.g. isocentric and fixed source-skin distance treatments, use of rectangular and blocked beams of different sizes and of wedges.

The basis for the procedure is the monitor calibration in the reference geometry: 10 cm depth in water, a source-detector distance equal to a) the isocentre distance (generally 100 cm) and a 10 cm x 10 cm field size at this distance or b) the regular source-skin distance (generally 100 cm) and a field size of 10 cm x 10 cm at this distance. The traditional monitor calibration referring to dose maximum has been replaced by a calibration at 10 cm depth for all beam geometries. The reason for this change is that the maximum dose depends on the degree of electron contamination which varies critically with change in beam geometry.

In order to simplify the procedure, measurements in a mini-phantom are recommended for several irradiation geometries. It is possible in this way to separate between the field-size dependence of the monitor response due to scatter in the accelerator (or ^{60}Co -unit) head and due to scatter in the water phantom.

Commercial treatment planning systems can often be used for monitor unit determinations for irradiation geometries used in clinical practice. It is, however, of great importance that these computations can be verified in an independent way to avoid errors and to improve accuracy. Furthermore, a uniform method used by different departments will help in communication between departments. It is also expected that common terms and concepts will simplify the publication of different dosimetry interaction coefficients and correction factors and make them useful to other departments. This will hopefully result in a database very much needed in this field, and ultimately in an improved consistency in dosimetry procedures in radiotherapy.

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1. INTRODUCTION

There is increasing evidence indicating that a high degree of accuracy in dose delivery is essential for the success of radiotherapy. From a review of dose-effect curves a standard requirement of 3.5% has been proposed for the accuracy needed in dosimetry. This figure is the total uncertainty (i.e. combined uncertainties of class A and B, see IAEA TRS 277) and is stated as one standard deviation in the dose value at the specification point or in the mean dose to the target volume (Wambersie et al., 1994).

A working group in the European Organisation of Research and Treatment of Cancer (EORTC) concluded that such a high accuracy in clinical dosimetry could be maintained if the following action levels were adopted (Johansson et al., 1986):

- $\pm 3\%$ for calibration of therapy units, and
- $\pm 5\%$ for the delivery of prescribed dose (when depth dose characteristics, shielding blocks, trays, wedges or other non-reference conditions are involved).

Even closer tolerance is desirable for some treatments (Brahme, 1988), particularly to distinguish results from different treatment techniques on patient groups, but would generally be unattainable in most routine clinical procedures. Work is, however, continually developing to improve the dosimetrical accuracy. The present document should be considered as part of this work.

The technical possibilities to generate optimised radiation fields have greatly improved, by such things as use of different field shapes, different beam qualities, wedges or other beam intensity modifiers. All these parameters influence the characteristics of the dose distributions, making accurate patient dosimetry more difficult. It is important therefore, to investigate and suggest methods that can be used for accurate dose determination for both simple and complex beam geometries.

Accurate dosimetry involves a large number of steps, starting with the calibration of a dosimeter through the determination of the absorbed dose at a point of interest in the patient. International and national protocols are available for the determination of the dose at a reference point in a standardised geometry. Subsequent steps are, however, not dealt with in such protocols. The use of common procedures is of importance to create a coherent system which should improve accuracy and allow meaningful comparisons of treatment results to be made, both inside a centre

and between centres. The adoption of common methods would also simplify external audits of the applied dosimetry procedures in a department.

The first step in the dosimetric chain is to base the ionisation chamber calibration on a common international standard. The BIPM (International Bureau of Weight and Measures) has developed international reference standards for ^{60}Co gamma radiation for the quantities air kerma and absorbed dose to water. The Primary or Secondary Standards Laboratories (PSDLs or SSDLs) compare or base their standards on calibrations at the BIPM. The calibration procedures and quality control programmes which guarantee acceptable accuracy for the calibrations have recently been outlined in a report from the IAEA (International Atomic Energy Agency) Report TRS 374 (IAEA, 1994).

Only a few PSDLs are able to calibrate dosimeters in terms of absorbed dose to water at beam qualities other than ^{60}Co gamma radiation. However, the IAEA has published a protocol, Report TRS 277, which includes conversion factors and procedures for using ^{60}Co gamma air-kerma calibrated chambers for determination of absorbed dose to water with photon beams from accelerators (IAEA, 1987). Various national protocols are also available many of which apply the same recommended data sets or are based on IAEA TRS 277.

The protocol described in TRS 277 deals with absorbed dose determination under reference conditions. These conditions are: usual treatment distance, a field size 10 cm x 10 cm at this distance, 5 cm depth in water to the effective point of measurement for ^{60}Co gamma radiation and high energy x-rays with quality index $T_{20/10} \leq 0.7$ (≈ 10 MV). For $T_{20/10} > 0.7$ a depth of 10 cm instead is recommended. The rationale for using two depths is that beams with low photon energies often are used for treatment of tumours situated at medium depth (i.e. ≈ 5 cm) while high energy photon beams mainly are used for treatment at greater depths. The purpose is to achieve the most accurate dose at the centre of the target volume. In addition, in order to ensure electron equilibrium at the point of measurement and to avoid the presence of contaminating electrons in the beam, the depth should be larger than the range of secondary electrons. It would, however, be more practical to use only one reference depth for all beam qualities in the calibration of an accelerator dose monitor. Moreover, the use of one reference depth would simplify the whole calculation procedure to determine the dose at the point of interest in a patient as one consistent formalism could be used. The IAEA TRS 277 could still be used as the basis for dosimetry, as values of dosimetric interaction coefficients reported for 5 cm

depth are also usable at 10 cm: the variation with depth (> 5 cm) being very small (Svensson and Brahme, 1986).

It has been common procedure to measure the dose at the reference depth and then to use relative depth dose information to establish the accelerator dose monitor calibration (output factor) valid for dose maximum. There are several problems in this procedure:

- The steps from absorbed dose measurement in the reference geometry to the determination of the dose at dose maximum at any field size (per accelerator monitor unit) are only briefly described in national or international protocols, or are omitted.
- This procedure must always include the use of depth dose data. The measurement will therefore include additional uncertainties as most detector systems have a different response at 5 or 10 cm from that at dose maximum. The response might depend on the field size as the electron contamination at dose maximum is a function of the collimator setting, use of blocks, wedges, etc.
- The output factor refers to different depths for a given accelerator potential as the depth of dose maximum varies with field size. These depths of dose maximum are not simple to determine using a detector with an extended sensitive volume (e.g. a cylindrical ionisation chamber), as the determination is critically dependent on the definition and use of an effective point of measurement.

In the present report it is strongly recommended that output factors are determined at 10 cm depth. The reference conditions should preferably be an isocentric set-up, generally a source - detector distance of 100 cm and a 10 cm x 10 cm field at this distance. Alternatively or in addition, an SSD of 100 cm could be used and a 10 cm x 10 cm field size at the phantom surface. In the present report it is thus assumed that the absorbed dose per accelerator monitor unit under these reference conditions, $\dot{D}_R$ has been determined according to well established recommendations (generally IAEA TRS 277). $\dot{D}_R$ should then be used as the basis for determination of absorbed dose at any point of interest along the central axis in a water phantom for various parameters used in radiation treatments.

Monitor units and complete dose distributions are to-day generally computed with treatment planning systems. The dose calculation algorithms and the dose computation procedures in such systems are often not completely known by the user. The accuracy in these computations must be tested by the local physics staff by

performing dose measurements in phantom for typical irradiation geometries. In the daily routine, the procedure described in this report could be applied to check dose calculations from a computer planning system at specific points. It is obvious that the accuracy in the present procedure ought to be of the order of one or two percent in order to determine the dose to the target volume with an uncertainty of less than 3.5 % (1 SD). Tests of the procedures indicate that this goal can be reached, if the uncertainty in dosimetry interaction coefficients from the dosimetry protocol (generally IAEA TRS 277) is disregarded.

The present report describes all the measurements and calculations needed for the determination of absorbed dose per accelerator monitor unit along the central axis. Common dosimetrical data for different accelerator units might sometimes be used to simplify the measurement procedures. A second document, currently under preparation, includes tables of such data.

The present report does not cover asymmetric fields, dynamic wedges and multileaf collimators. Procedures for those are, however, under preparation based on the suggested formalism contained in this report.

2. LIST OF SYMBOLS

The symbols for the physical quantities within this report are those generally adopted by ICRU, in which one single letter is used instead of several letters as sometimes used in the literature, e.g. TPR_{10}^{20} is replaced by $T_{20/10}$. The symbol $\dot{D}$ is used by ICRU to indicate dose per time interval. In the present report, $\dot{D}_R$ indicates for ^{60}Co units the dose per time interval measured at the reference geometry. In addition $\dot{D}_R$ is used as the symbol for the dose per accelerator monitor unit also measured at the reference geometry.

There is a trend from fixed source-skin distance treatments to isocentric treatments. When a fixed source-skin distance is used this has therefore been indicated with an apostrophe (').

2.1 SYMBOLS

B	fraction of the prescribed dose delivered with a wedge in the beam
c	treatment field size defined by the collimator at the reference distance f_R or f'_R
c_e	collimator equivalent square for a rectangular collimator setting (X,Y), where X and Y are the openings of the upper and lower jaws, respectively. The collimator equivalent square should be used for the quantities related to the head scatter factor
c_R	reference field size defined by the collimator at f_R or f'_R . A 10 cm $\times$ 10 cm field size is recommended
c'_R	reference field size defined by the collimator at $f = f'_R + z_R$. A field size $c'_R = c_R \cdot \frac{f'_R + z_R}{f'_R}$ is recommended : i.e. 11 cm $\times$ 11 cm if $f'_R = 100$ cm
D	absorbed dose in treatment conditions at the reference distance f_R
D_0	absorbed dose measured in the mini-phantom
$D_p(z)$	Primary component of the absorbed dose at depth z in a phantom

D_T	dose prescribed to a patient
D'	absorbed dose in treatment conditions at the reference source-skin distance f'_R
D'_T	absorbed dose in treatment conditions at the source-skin distance $f'_T \neq f'_R$
$D'_{20/10}$	ratio of percentage depth dose at 20 cm and 10 cm depth, at fixed source-skin distance, and a field size 10 cm $\times$ 10 cm at the phantom surface
$\dot{D}_R$	absorbed dose per monitor unit, under isocentric conditions determined in a large water phantom in reference conditions of depth (z_R) and field size (c_R) at a source to point of interest distance f_R
$\dot{D}'_R$	absorbed dose per monitor unit, under fixed source-skin distance conditions determined in a large water phantom in reference conditions of depth (z_R) and field size (c_R) at a source-skin distance f'_R
$\dot{D}'_T$	absorbed dose per monitor unit, at a fixed source-skin distance f'_T , determined in a large water phantom in reference conditions of depth (z_R) and field size (c_R)
e_w	ratio of the effective attenuation coefficients of the open and wedged fields
f	distance from source to point of interest
f_R	reference distance from source to isocentre (= distance from source to point of interest under isocentric conditions)
f_T	treatment distance from source to skin (when differing from f_R)
f'	fixed source-skin distance
f'_R	reference distance from source to skin under fixed source-skin distance conditions.
f'_T	treatment distance from source to skin (when differing from f'_R)

$k_{o,b}(c, s_b)$	correction factor for the presence of shielding blocks in the beam determined with the mini-phantom at reference depth z_R (isocentric conditions); c and s_b represent the field sizes defined by the collimator and by the shielding blocks respectively at f_R
$k_b(c, s_b)$	correction factor for the presence of shielding blocks in the beam determined in a large water phantom at reference depth z_R (isocentric conditions); c and s_b represent the field sizes defined by the collimator and by the shielding blocks respectively at f_R
$k'_b(c, s_b)$	correction factor for the presence of shielding blocks in the beam determined in a large water phantom at reference depth z_R (fixed source-skin distance conditions); c and s_b represent the field sizes defined by the collimator and by the shielding blocks, respectively, at f'_R
$k_{o,t}(c)$	transmission factor of the shadow tray supporting the shielding blocks determined at z_R with the mini-phantom for the field size c at f_R (isocentric conditions)
$k_t(c)$	transmission factor of the shadow tray supporting the shielding blocks determined at z_R in a large water phantom for the field size c at f_R (isocentric conditions)
$k'_{o,t}(c)$	transmission factor of the shadow tray supporting the shielding blocks determined with the mini-phantom for the field size c at f'_R (fixed source-skin distance conditions)
$k'_t(c)$	transmission factor of the shadow tray supporting the shielding blocks determined at z_R in a large water phantom for the field size c at f'_R (fixed source-skin distance conditions)
$k_{o,w}(z_R, c_R)$	wedge factor determined in the mini-phantom at depth z_R for the field size c_R at f_R (isocentric conditions)
$k_w(z_R, c_R)$	wedge factor determined in a large water phantom at depth z_R for the field size c_R at f_R (isocentric conditions)
$k'_{o,w}(z_R, c_R)$	wedge factor determined in a mini-phantom at depth z_R for the field size c_R at f'_R (fixed source-skin distance conditions)

$k'_w(z_R, c_R)$	wedge factor determined in a large water phantom at depth z_R for the field size c_R at f'_R (fixed source-skin distance conditions)
$O_0(c)$	output ratio determined with the mini-phantom for the field size c at f_R (isocentric conditions)
$O_R(c)$	output ratio determined at the reference depth z_R in full scatter conditions for the field size c at f_R (isocentric conditions)
$O'_0(c)$	output ratio determined with the mini-phantom for the field size c at f'_R (fixed source-skin distance conditions)
$O'_R(c)$	output ratio determined at the reference depth z_R in full scatter conditions for the field size c at f'_R (fixed source-skin distance conditions)
$O'_T(s)$	output ratio determined at the reference depth z_R in full scatter conditions for the field size s at the source-skin distance f'_T ($f'_T \neq f'_R$)
$P(z, s, f')$	percentage depth dose at depth z and for a square field size s : ratio of the dose at a given depth z to the dose at the depth of maximum dose z_m in a water equivalent phantom, for the same field size s at the same source-skin distance f' .
$P_R(z, s, f'_R)$	reference percentage depth dose at depth z and for a square field size s : ratio of the dose at a given depth z to the dose at the reference depth z_R (10 cm) in a water equivalent phantom for the same field size s at the same source-skin distance f'_R .
r	radius of a sector in the Clarkson's method
R	equivalent radius of a square or rectangular field : radius of a circular field having the same percentage depth dose as a square or rectangular field
s	treatment field size defined by the collimator at f'_T in fixed source-skin distance conditions
s'	treatment field size defined by the collimator at $f'_T + z$
s''	treatment field size defined by the collimator at $f'_T + z_R$

s_b	treatment field size defined by the shielding blocks at the point of interest in isocentric conditions (f_R) or at the skin in fixed source-skin distance conditions (f'_R or f'_T) (for blocked fields)
s'_b	treatment field size defined by the shielding blocks at $f'_T + z$
s''_b	treatment field size defined by the shielding blocks at $f'_T + z_R$
s_e	equivalent square of a rectangular field. s_e should be used for the quantities related to phantom scatter
s_R	reference field size defined by the collimator at f'_T
$S(z,c)$	scatter-primary ratio defined as the ratio of the scatter contribution of absorbed dose to the primary contribution of absorbed dose, at depth z in a phantom for a field size c at that depth.
$S_0(z)$	scatter-primary ratio defined as the ratio of the scatter contribution of absorbed dose to the primary contribution of absorbed dose, at depth z in a mini-phantom .
$S(r,\theta)$	scatter-primary ratio at a point of polar coordinates r and θ
$T(z,s)$	tissue-phantom ratio at depth z for a square field size s at the detector distance: ratio of the absorbed dose at depth z to that at reference depth z_R measured in a large water phantom at a fixed source-detector distance
$(T_0)_{z/z_R}$	ratio of ionisation chamber readings at depth of z and z_R , at constant source detector distance, measured in a mini-phantom
$T_{20/10}$	ratio of absorbed doses at depths of 20cm and 10cm, measured in a large water phantom, at constant source-detector distance for the reference field size 10 cm $\times$ 10 cm. This quantity is also called the quality index (when $z_R = 10$ cm, $z = 20$ cm and $s = 10$ cm $\times$ 10 cm, $T(z,s) = T_{20/10}$)
$(T_0)_{20/10}$	ratio of ionisation chamber readings at depth of 20cm and 10cm, at constant source-detector distance, measured in a mini-phantom
U	number of monitor units

$V(z,s)$	volume-scatter ratio; ratio of the dose at depth z in a large water phantom to the dose in the mini-phantom for the same field size s and the same source-point of interest distance
z	treatment depth
z_m	depth of maximum dose
z_R	reference depth
μ	linear attenuation coefficient
θ_E	effective wedge angle
θ_i	sector angle
θ_w	nominal wedge angle
Ψ	energy fluence

The definitions given for O_R , O_O , V , P_R , T etc., refer to open fields. When the letter w appears into brackets (for example $O_R(c,w)$ or $T(z, c, w)$), the quoted quantity refers to a wedged beam and its definition is similar to that given for the corresponding quantity used for an open beam. When the index b is used for P_R and T ($P_{R,b}$, T_b), those quantities refer to a blocked beam with a shadow tray in the beam and their definitions are similar to those given for the corresponding quantities used for open beams.

3. RADIOTHERAPY EQUIPMENT

3.1 INTRODUCTION

This chapter deals with aspects of the design of radiotherapy equipment which affect the beam characteristics, and which are therefore important when performing monitor unit calculations for high energy photon beams.

3.2 THE RADIATION SOURCE

3.2.1 ^{60}Co Gamma-ray Source

In a ^{60}Co unit, gamma rays are produced by the disintegration of ^{60}Co nuclei. For each disintegration event, two gamma rays are produced with energies of 1.17 and 1.33 MeV. These are accompanied by the emission of a beta particle (electron) which is absorbed within the source or its container and which can be effectively ignored.

The source consists of either pellets or small cylinders of cobalt within a welded steel container. The source is usually a cylinder of 1-2.5 cm diameter and 2-2.5 cm height. Emission from the source is approximately isotropic, i.e. the same in all directions, and this emission pattern is modified by the source housing and collimator system. A significant number of emitted photons is scattered by this housing and collimation system, generating lower energy photons which contribute 30% of the fluence to the treatment beam, as can be derived from the energy spectrum of figure 3-1.

At large distances, the source of radiation is effectively a point. Consequently the photon fluence decreases with the inverse-square of the distance from this source. This causes a small variation of fluence in a plane perpendicular to the central axis of the beam, which complements a greater variation caused by a decrease in fluence of scattered photons with distance from the central axis.

3.2.2 Accelerator X-Ray Source.

With accelerator produced photon beams, the radiation is produced when a pulsed stream of high-energy electrons impinges upon a target. This target is a disc of metal of high atomic number which is mounted at the end of the evacuated wave guide along which the electron stream is flowing. When the electrons are suddenly stopped in the target, bremsstrahlung photons (x-rays) are produced.

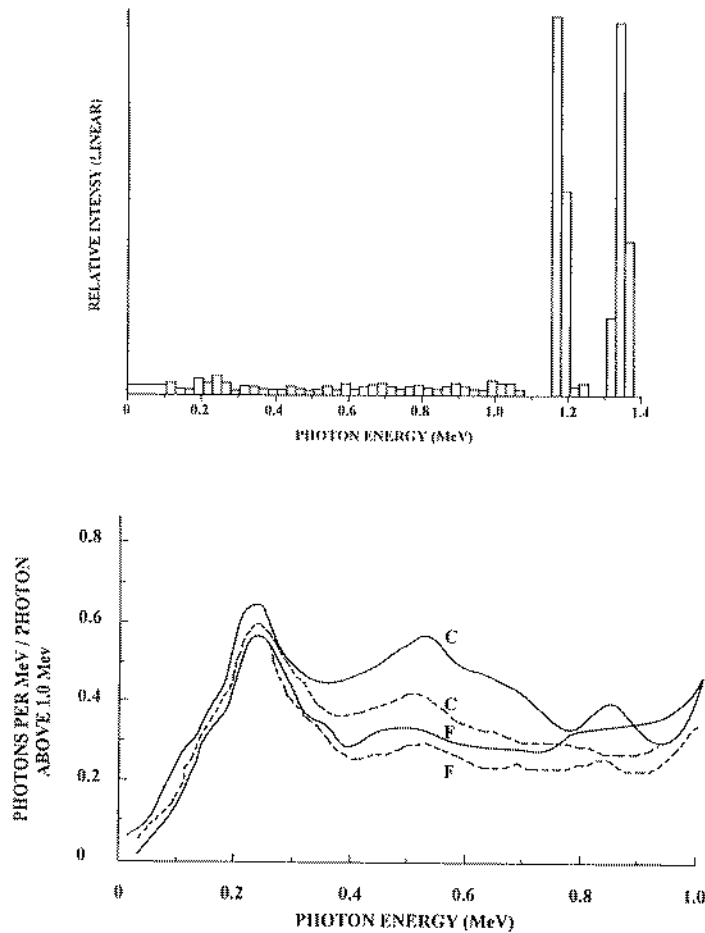


Figure 3-1 : Energy spectra of ^{60}Co beams from radiotherapy units a) From Jessen (1973) and b) from Löfroth et al., (1973). Data are given for $6 \times 6 \text{ cm}^2$ and $15 \times 15 \text{ cm}^2$ field sizes for two different types of ^{60}Co units (SSD 100 cm). It is seen that there are approximately 0.4 to 0.5 photons below 1 MeV for each photon above 1 MeV.

The cross sectional size of the stream of electrons at the target is typically only a few millimetres, and therefore the 'source' of x-rays is small. The x-ray fluence varies with both distance and angle of emission. The distance dependence of the fluence is primarily determined by the inverse-square law. The latter dependence arises because the angular emission of x-rays is very non-isotropic: at high energies, the Bremsstrahlung photons are emitted preferentially in the direction of travel of the incident electrons - a feature which becomes more pronounced for higher incident electron energies.

X-rays are emitted with a continuous distribution of energies up to the maximum energy corresponding to that of the incident electrons, determined by the accelerating system conditions (figure 3-2).

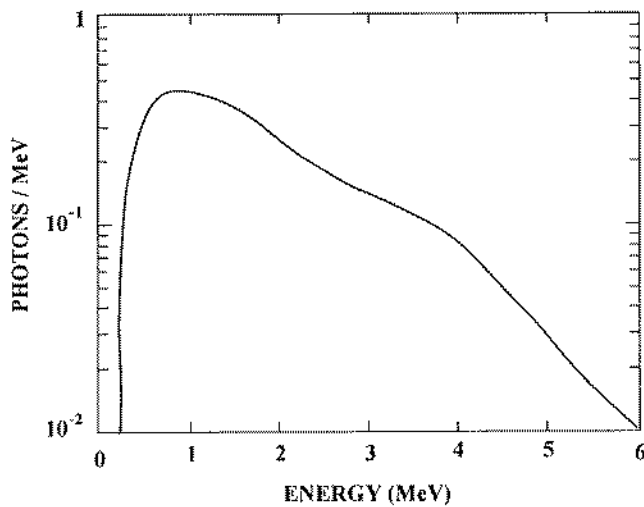


Figure 3-2 : Photon energy spectrum of a 6 MV x-ray beam from Mohan and Chui (1985). The curve is redrawn and is based on Monte Carlo calculations. SSD = 100 cm, computation for a radial range of 0-3 cm. The mean energy of the photons is 1.91 MeV.

3.2.3 Beam Flattening and Hardening.

The x-ray beam produced in the target must be modified to make it suitable for therapy:

- * by removal of low-energy photons which would be more readily absorbed in tissues which overlay the target volume;
- * by modifying the intensity distribution in order to produce a uniform dose profile at depth.

These modifications are achieved by the use of filters.

The unperturbed fluence profile from the target of a linear accelerator is peaked along the central axis of the beam, due to the predominantly forward production of photons in the target. A beam flattening filter is used to generate the desired primary fluence profile for the beam. Such a profile is purposefully decreased on the central beam axis to compensate for the reduction in phantom-scattered

radiation, at the beam edges. The flattening filter is therefore roughly conical in shape, with the thickest part aligned with the central axis. For cobalt units, the unperturbed primary fluence profile is much more uniform, so that flattening filters are not generally employed.

The flattening filter will affect the x-ray spectrum by selectively absorbing low energy x-rays, and by producing scattered photons which are then incident upon the patient. The effects of beam flattening, beam hardening and beam scattering are therefore interrelated and the design of a suitable filter is complex. For low-energy machines, beam hardening is best achieved using high atomic number materials. For the highest energy machines, such materials would remove high energy photons by means of pair-production interactions, and in these situations beam hardening can best be achieved using materials of medium or low atomic number.

The scattered photons produced by the flattening filter may contribute up to 10% of the total dose to the patient, and since these photons are scattered through small angles and with little change in energy, they have similar penetration characteristics as the unscattered component.

Some accelerators use alternative methods for beam flattening such as electron-beam scanning. Such accelerators may need to incorporate a filter specifically for beam hardening.

3.3 COLLIMATION

Several collimators are used to define the beam. Close to the radiation source is a block of material of high density, high atomic number, which has a conical aperture. This block, together with other shielding, forms the primary collimator, and restricts the beam size to slightly greater than the maximum size used for treatment. Beyond this block are the secondary collimator blocks which are moved in and out to produce rectangular field shapes. The detailed positioning and design of the blocks can affect the size of the beam penumbra. In some designs the penumbra is minimised by ensuring that the inner surface of the blocks always projects back to the radiation source for all possible openings. Other designs include non-flat inner surfaces which ensure consistency of penumbra.

The collimators may be coupled together in X and Y pairs (lower and upper pairs, respectively), so that opposing blocks assume identical displacements from the central axis (the axis of rotation of the collimator system, which coincides with the line from the isocentre to the radiation source). Independent movement is now becoming increasingly common, as this allows the generation of rectangular beams

which are offset from the central axis (asymmetric beams). It is important therefore to distinguish between the **central axis** as defined above, and the **beam axis** which represents the projection of the centre of a defined collimator opening. For symmetric beams these are coincident, but they differ for asymmetric beams. It should be noted that in this report we are only dealing with monitor unit determination at points along the central axis. The **principal planes** of a rectangular beam are the planes which contain the **central axis** and are parallel to one side of the rectangle.

Multileaf collimators are also becoming readily available. A multileaf collimator system consists of a pair of arrays of adjacent collimating "leaves" (i.e. thin blocks), where each leaf can be moved independently. This allows the generation of irregular, asymmetric field shapes. Such a system may replace one of the conventional collimator pairs, or may be additional. In the latter case, the distance between patient and collimator will be reduced.

It is necessary to distinguish between the terms collimator setting and field size. The **collimator setting** is the nominal setting of the collimator, usually a scale readout which indicates the beam dimensions at the isocentre. The **field size** indicates the beam dimensions at some distance of interest which may differ from that of the isocentre. The field size therefore equates to the collimator setting at the isocentre distance, for unblocked beams.

3.4 THE BEAM MONITOR

The photon fluence is monitored by a flat ionisation chamber that extends across the opening in the primary collimator, and is positioned (nowadays) between the flattening filter and the secondary collimator. The ionisation chamber consists of several elements, the signals from which are used to control a number of beam parameters in addition to radiation dose, such as symmetry and uniformity of the radiation beam. Accelerators have two independent chamber systems to monitor the delivered radiation - a primary and a secondary (back-up) channel. Because the radiation from a ^{60}Co unit is entirely predictable, these units have a timer system in place of ionisation chambers to control the delivered radiation.

The primary objective in "calibrating" this monitor is the determination of the relationship between the signal from the beam monitor system (expressed in monitor units) and the absorbed dose delivered under reference conditions. This procedure is described in detail in the IAEA protocol TRS 277 (IAEA, 1987).

The position of the monitor chamber in the treatment head is an important consideration. The ionisation current from the chamber may be enhanced by

backscattered photons and electrons from the collimators, which may vary considerably with collimator opening (Kubo 1989, Luxton and Astrahan 1988, Lam et al. 1996). This may lead to increased variation in dose per monitor unit with changes in field size. Furthermore, differences in backscatter received by different parts of the monitor chamber may affect the control of the beam parameters. To minimise these effects, the monitor chamber may be protected from changes in backscatter by the inclusion of a metal plate between the chamber and the collimator assembly, and by maximising the separation between monitor chamber and collimators.

Should the accelerator change its basic operating parameters, the various beam characteristics will also change, leading to variations in profile shape and beam quality. The relationship between monitor units and dose under reference conditions may also change. A regular quality control programme may identify such changes. In addition, the linearity of dose with monitor setting can be affected by changes which occur in the operational parameters during a single exposure. It is therefore necessary to establish the dose per monitor unit relationship for a number of different monitor settings.

3.5 BEAM MODIFIERS

3.5.1 Wedge Filters

The purpose of the wedge filter is to modify the dose distribution in one of the principal planes of the beam so that the isodose lines are rotated relative to their open beam position. This can be done by inserting a metal wedge in the beam, hence the name. The detailed shape of a wedge filter may be complex to produce the desired dose distribution.

Use of a wedge will produce additional scatter which will increase with wedge thickness. The relative amount of scattered radiation which reaches the patient will depend upon the wedge position and field size (Zhu et al., 1995).

Wedges may be mounted on the face-plate of the head, and are termed **external wedge**. Alternatively, the wedge may be placed inside the head upstream of the adjustable collimator (**internal wedge**). Such a wedge can be moved in or out of the beam by remote control. Since there is only room for one wedge mounted internally, it is usual to incorporate a wedge with a steep wedge angle (e.g. 60°) and produce shallower angles by mixing wedged and open beams in suitable proportions. This is called a **universal wedge**. The wedge can be designed for large fields, but the larger the field covered, the thicker the wedge has to be, which has dosimetric consequences.

One further technique to produce wedge-shaped dose distributions requires the movement of one collimator progressively across the field during irradiation. The velocity may be programmed to give a dose profile similar to that produced by any conventional wedge. This is a **dynamic wedge**.

3.5.2 Blocks and Trays

For some treatments, additional blocks of a metal or alloy of high density may be used to shape the field to conform to the target volume or to protect organs at risk (multileaf collimators may be used for the same purpose). Optimal block placement is achieved using blocks which are formed with surfaces that project back to the source, so-called "focused" or "divergent" blocks.

The tray on which the blocks are mounted attenuates the beam slightly. The use of blocks also changes the scattered radiation component of the beam by :

- * Producing additional photon and electron scatter from the block tray. Electrons ejected from the block tray will increase the dose close to the surface. This effect is strongly dependent upon the distance between the tray and the patient surface.
- * Modifying the volume in the patient from which scattered photons are generated. This will change the dose also at the beam axis.
- * Partially shielding the head scatter.

3.5.3 Compensators

When the surface of the patient is very uneven as in the head-and-neck region, it can be difficult to achieve a uniform dose over the target volume, since the beam is attenuated and scattered differently in different regions. A compensator may be used to counter-balance these effects by interposing various thicknesses of material in different parts of the beam. Compensators may also be used to modify dose distributions which are adversely affected by internal body structures (e.g. air or bone).

The use of compensators will affect the calculation of dose per monitor unit in a manner similar to wedges.

3.5.4 Spoilers and Bolus

On occasion, it may be desired to increase the superficial dose for high-energy photon beams which show significant "skin-sparing". The most common technique is to place some water-equivalent material on the patient's skin, i.e. **bolus**. The dosimetric consequences are handled by treating this bolus as part of the patient, increasing the treatment depth and decreasing the source-surface distance by the bolus thickness.

The skin dose can also be increased by adding a **spoiler** to the beam, essentially a block tray without blocks, the purpose of which is to produce secondary electrons which travel on to the patient surface.

4. PHYSICAL AND DOSIMETRIC BEAM CHARACTERISTICS

4.1 INTRODUCTION

When a beam of photons enters a patient (or phantom), energy is transferred from the beam and deposited within the patient as a result of photon interactions.

This chapter concentrates on this aspect and introduces various concepts, models, terms and physical processes which relate to the pattern and magnitude of this energy transfer. General characteristics of these processes are discussed, together with their relative importance.

4.2 PRIMARY AND SCATTERED COMPONENTS

One basic physical principle which has been used for many years in photon dose calculations involves partitioning the dose delivered to each volume element into two components - **primary** and **scatter**. The photons that enter through the surface of the patient or phantom are referred to as **incident** or **primary photons**. The first time such photons interact within the medium, they contribute to **primary dose** through secondary electrons (see section 4.3.3). **Scattered component** is that part of the dose which is deposited by photons which have previously interacted at least once in the medium.

These two components obey the same physical laws (divergence, attenuation, generation of secondary photons and electrons), but have very different source distributions and energy spectra. Hence, mathematical expressions derived to describe their effects, whether exact or by approximation, have little in common.

Dose calculation models which achieve separation of these components have gained considerable success when calculating complex treatment plans.

4.3 PHOTON AND ELECTRON TRANSPORT

4.3.1 Secondary Electrons

A second fundamental distinction differentiates between the primary and scattered photons, and the movement of the electrons generated by these photons. This distinction becomes increasingly important for higher energy photons. When photon energy is low, the range of the electrons produced in the medium is very short and a point deposition approximation (i.e. energy of electrons is deposited where they

are produced) is reasonable. The range of the most energetic secondary electrons will reach 1 mm at about 500 keV but 10 mm at about 2 MeV. For high energy machines, point deposition is no longer applicable and electron transport must be taken into account for accurate calculation of dose.

4.3.2 Kerma and Dose.

At low photon energies, where point deposition of energy may be assumed, collision kerma equals dose, i.e. the kinetic energy released per unit mass (ignoring subsequent bremsstrahlung losses) equals the energy absorbed per unit mass. This is a special, simple case of **electron equilibrium**. For higher energies, electron transport cannot be ignored. In these circumstances, electron equilibrium exists when the sum of all energy carried away by electrons (produced by interacting photons) from a small volume equals the sum of all energy carried into that region by electrons generated from the surroundings. Lack of equilibrium occurs where the production of electrons rapidly changes, such as near beam edges and near external surfaces. The region near the entrance surface of the beam is the most obvious example. Here, there is a greater transfer of radiation energy to kinetic energy than is absorbed in the medium i.e. collision kerma exceeds dose. Lack of equilibrium, although to a much smaller extent than exists at the surface, arises from the gradient introduced by the attenuation and divergence of photons in a medium. Sometimes, this condition is referred to as "**transient electron equilibrium**".

4.3.3 Primary Dose

Those photons which are incident upon the surface of the patient (or phantom) are termed **primary**. Some of those will have travelled directly from the radiation source: others will have been produced as a result of interactions within the structures comprising the treatment head. Most of these interactions are Compton-scattering events, which generate secondary electrons in addition to scattered photons (see 4.3.7). In addition, pair production can occur if the incident photon has sufficient energy (i.e. above 1.02 MeV), resulting in the generation of electron-positron pairs and subsequent generation of photons of 0.51 MeV from positron annihilation. The incident or primary radiation therefore consists of photons coming directly from the source, together with photons and electrons produced within the beam by the above interaction events. Since these occur mostly in structures in the head of the machine (some occur in the intervening air), we refer to the photon component as head-scattered photons, but they still belong to the "incident" or "primary" group.

The dose from primary photons to the central axis of an open beam at depth in a phantom will depend upon:

- * Distance from the source (f)
- * Head-scatter, which will depend upon collimator setting (c) and other beam defining parts (blocks, wedges)
- * Attenuation, i.e. absorption and scattering in the phantom, determined by depth (z) and beam energy (E).

Electron transport and the lack of electron equilibrium will in addition determine the local energy deposition, i.e. the dose.

At any depth, the dose contribution from primary photons is determined by the collimator setting, and not by the size of the radiation field at that depth.

4.3.4 Electron Contamination

The primary photon beam is contaminated with electrons generated in the flattening filter, collimator, air and other intervening material in the beam between the accelerator and the patient. For an open beam (i.e. without wedge, tray and blocks) at SSD 100 cm and an accelerating potential up to about 10 MV the electrons generated in air generally dominate. At higher potentials instead, electrons from the flattening filter are of greater importance (Nilsson, 1985). Figure 4-1 shows an experimental determination of the electron contamination. Clean open beams were obtained using magnetic deflection of the contaminating electrons leaving the accelerator head. Addition of different materials in the beam influences widely the dose distribution at depths up to the range of these electrons. From Figure 4-1 it is seen that such electrons are of importance up to about 6 cm for 21 MV x-rays. The contaminating electrons complicate dose measurements and the reference depth for dose determination should be beyond the range of these electrons. Therefore 10 cm is recommended as reference depth in this report which should exclude contaminating electrons up to at least 40 MV.

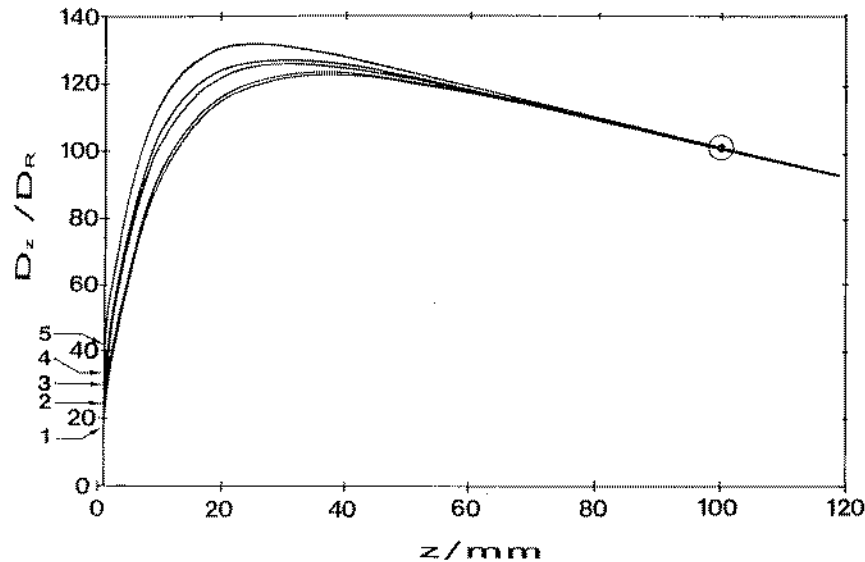


Figure 4-1. Depth dose curves for a 21 MV photon beam. Curve n° 1 is obtained using magnetic deflection of contaminating electrons. Curve n° 2 is for a beam without magnetic deflection. Curves 3-5 are for some typical filters and blocks introduced in the beam in daily clinical practice, with no magnetic deflection. SSD and field size are the same in all cases (SSD = 100 cm and field size 7 cm x 7 cm) (personal communication R. Sjögren, Umeå).

4.3.5 Inverse-Square Law

The number of photons per unit area from a point source decreases rapidly with the distance from the source. In vacuum, it follows the "inverse-square law", i.e. the fluence is inversely proportional to the square of the distance. For a high-energy photon beam, this is not exactly correct, since the head-scattered photons come from several components at various distances from the target. The total primary fluence therefore behaves as if the source was slightly displaced (the "virtual" source) (McKenzie and Stevens 1993). The effect needs to be considered only when distances are different from those at which data were measured or calibrations made (i.e. SSD vs. isocentric), especially when a wedge of large angle is in the beam. An alternative to the "virtual source" approach is to use a two-source model, in which a small "true" x-ray source is considered together with an extended source of the head-scattered photons (Ahnesjö et al., 1992, Dunscombe and Nieminen, 1992, Jaffray et al., 1993).

In this report, it is assumed that the inverse-square law holds.

4.3.6 Effects of Wedges and Blocks

If a wedge is inserted in the beam, the fluence distribution of primary photons is modified. In addition, head-scattered photons are generated. This effect is larger for internal than external wedges. This increase in head-scatter is the main reason why wedge factors vary with field size.

Blocks introduced into the beam may modify the head-scatter by partly obscuring the structures inside the accelerator head that generate such photons. There may in addition be a small amount of scatter generated, particularly at low photon energies (Lam and Ten Haken, 1996). For points on the central axis, these effects are small, except for the situations where the block is very close to, or obscures, the central axis (Van Dam et al., 1992, Thatcher and Bjärngård, 1994).

The additional photon fluence due to head-scatter is generally considered to be the same throughout the beam as in its centre. This is an approximation, and with the increased utilisation of asymmetric beams and beams shaped by multileaf collimators, techniques to account for its variation are being developed (e.g. Chaney et al., 1994, Lam et al., 1996, Yu and Sloboda, 1996).

4.3.7 Phantom Scattered Photons

When the primary photons interact in the irradiated medium, they do so mostly by Compton scattering which dominates in materials of low atomic numbers in the energy range mainly used in radiation oncology. When an incident photon interacts by a Compton type scatter, a secondary photon is produced. At high energies, the secondary photons tend to be emitted in the forward direction with little energy loss. The secondary photons can, in their turn, produce other scattered photons.

In treatment planning systems, some dose calculation algorithms distinguish also between once-scattered (i.e. scattered photons produced by a primary photon) and multiple-scattered photons (scattered photons produced by interaction of scattered photons).

The proportion of dose due to scattered photons depends upon photon energy and the volume of patient irradiated (field size, depth and patient thickness).

4.3.8 Dose Profiles

The primary photon fluence is not constant across the beam. In the "open" beam, the flattening filter is introduced to make the fluence uniform but some

variation (see section 3.2.3) still exists; and in a wedged beam the fluence is deliberately and very significantly distorted. When calculating doses, the variation in primary contribution may be accounted for by use of primary off-axis ratios, i.e. the ratio of the primary contribution off-axis to that on the central axis. Calculation of scatter contribution, however, requires integration over the entire field, with the primary off-axis ratios used as factors weighting scatter production.

Some empirical dose calculation methods use off-axis ratios in a combined (primary and scatter) way, such that these total off-axis ratios must vary with both field size and depth.

4.4 BEAM QUALITY

4.4.1 Beam Quality Specification

Beam production involves both generation and collimation processes. The sections on the radiation source (3.2) and on photon and electron transport (4.3 above), describe those processes which determine the resultant energy spectrum of the beam. These processes are complex, and depend upon the geometrical design and construction of the treatment head. As a result, beams generated by different equipment operating at the same nominal energy may vary significantly in their spectral distributions. Because low-energy photons are generally more readily attenuated in tissue than are high-energy photons, statement of the nominal energy of a beam is a poor specifier of the resultant beam and its properties.

As a result, beam quality is usually specified in terms of the penetrating properties of the beam. Various specifications have been used for this, based usually upon the relative doses at two specified depths in tissue (water) under reference conditions. These include:

* $D'_{20/10}$ The ratio of absorbed dose at 20 cm and 10 cm depth, at fixed SSD and a field size 10 cm $\times$ 10 cm at the phantom surface. These two measurements may be performed simultaneously, as required in the IAEA postal TLD measurement of beam quality (NACP 1980, Cunningham and Schultz 1984).

* $T_{20/10}$ The ratio of absorbed dose at depths of 20 cm and 10 cm, at constant source-detector distance and a field size 10 cm $\times$ 10 cm at the effective depth of the detector. This quantity is referred to within this text as the **quality index**. This measurement is frequently used within dosimetry protocols (AAPM 1983, Bridier et al., 1986, IAEA 1987).

* $(T_0)_{20/10}$ The ratio of ionisation chamber readings at depth of 20 cm and 10 cm, at constant source-detector distance, but measured in a mini-phantom. This measurement is particularly useful in assessing the variations in beam quality across a beam (Nyström and Karlsson 1994, Zefkili et al. 1994, Georg and Dutreix 1997).

The depth of 80% dose level relative to the dose maximum is also sometimes used. This method of specification is influenced by electron contamination at the depth of dose maximum, as discussed in section 4.5.2, and is therefore not recommended.

Within this report, beam quality will be expressed in terms of quality index for those parameters which are critically dependent upon quality, and in more general terms (e.g. nominal MV) where only a general indication of quality is necessary. Comparison of some of the above quality specifiers may be found elsewhere (Dutreix and Bridier 1985).

4.4.2 Attenuation and Beam Hardening

The probability that a primary photon will reach a point at depth z is dependent upon the likelihood of interactions with the medium it traverses. Attenuation of the primary photons in water is equal to $e^{-\mu z}$, where μ is the average linear attenuation coefficient of water for the photons in the beam. Since the spectrum is broad, photons of lower energy will be attenuated more than higher-energy photons, such that the beam becomes steadily more penetrating. Thus, μ may be reduced with depth. This is termed "**beam hardening**".

The value of μ is also likely to change when the beam has to pass through a wedge filter. Removal of low-energy photons will result in the beam being "harder", i.e. more penetrating. At very high photon energies, a sufficient number of the highest-energy photons may be removed by pair-production in the high-atomic-number wedge material accompanied by 0.51 MeV photons from positron annihilation, so that the beam will be slightly less penetrating, i.e. "softer".

Since photons in different parts of the beam are generated at different angles within the target and pass through different thicknesses of the flattening filter (and also of the wedge, if one is used), one would expect noticeable variations of the attenuation coefficient μ within the beam. The variation may be significant when calculating doses for points at some distance from the central axis, especially when a wedge of large angle is used.

4.5 DOSIMETRIC QUANTITIES

4.5.1 Definitions and Nomenclature

The following overview describes the quantities used within this text, indicating the nomenclature used.

Percentage Depth Dose.

Central axis percentage depth dose is the percentage ratio of the dose at depth z on the central axis of the beam to the dose at z_m , the depth of maximum dose on the central axis. These doses are determined for constant SSD and field size at the surface.

Tables of typical percentage depth doses have been produced for various beam qualities and for various field sizes (BJR, 1983).

For any given beam quality, percentage depth dose is a function of depth z , field size (s) at the source-surface distance (f'). The nomenclature used in this text for this quantity is $P(z,s,f')$ (e.g. see chapters 5 and 6).

Tissue-Air Ratio

The tissue-air ratio is the ratio of the dose at depth in tissue to the dose in air (i.e. to a small volume of tissue situated in air and surrounded by just enough material to generate electron equilibrium) at the same distance from the source and the same field size.

This concept was widely used in calculations of isocentric treatments using low-energy radiation beams, but has inherent problems at higher energies where the volume necessary for the material surrounding the point in air becomes large, such that consideration of a point in air loses validity. However, because it is conceptually simple, various attempts have been made to redefine the quantity (BJR, 1983, Khan, 1984).

This report does not use this quantity because of these difficulties and uncertainties but has introduced a related quantity : the volume scatter ratio, V .

Tissue-Phantom Ratio

The tissue-phantom ratio is the ratio of the dose at depth z to that at reference depth z_R in a water equivalent phantom for the same collimator setting and source-detector distance.

For a given beam quality, the tissue-phantom ratio is a function of depth z , field size s at depth z , and reference depth z_R . The nomenclature used within this report is $T(z,s)$ and the reference depth z_R is 10 cm.

From the definition of the quality index (see section 4.4.1), it follows that two beams of the same quality index have the same attenuation and scattering characteristics and consequently identical values of the tissue-phantom ratio for z larger than z_R (Bridier et al., 1986). However, when z is smaller than z_R , the contribution of contaminating electrons can modify the T value at depth as large as the maximum range of secondary electrons, i.e. at depth much larger than the depth z_m of maximum dose. The higher the quality index, the larger the variation of T with the head design and with the presence of beam modifiers such as trays or external wedges (a review of the relevant literature can be found in Beauvais et al., 1993 and Ahnesjö et al., 1989). It is therefore necessary to measure T carefully for each accelerator head design, for z values from 0 to the depth of z_R .

Output Ratios.

The output ratio in water $O_R(c)$ is the ratio of the absorbed dose at the reference depth for field size c , to the dose at the same depth for the reference field size c_R , measured in a large water phantom, where both c and c_R are defined at the reference distance f_R , i.e.:

$$O_R(c) = \frac{D(z_R, c)}{D(z_R, c_R)} \quad (4.1)$$

where these dose values are determined at the isocentre. The output ratio in water $O'_R(c)$ is similarly defined, but under conditions of fixed SSD and with the field size defined at the surface.

The ratio $O_O(c)$ is defined as the ratio of the absorbed dose at the reference depth for field 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 . Similarly an output ratio $O'_O(c)$ can be defined under conditions of fixed SSD and with the field size defined at the surface.

Tissue-Output Ratio.

The tissue-output ratio is the ratio of the dose at depth (z) for field size (s) to the dose at the reference depth (z_R) for a reference field size (c_R), at the same source-detector distance.

This ratio is equivalent to the product of the tissue-phantom ratio $T(z,s)$ for depth z and field size s , and the output ratio $O_R(s)$. This quantity may be used to simplify the expressions derived for monitor unit calculation in Chapter 5, and has been used by other authors in deriving similar relationships. However, it includes elements of both phantom scatter and head scatter, and because these components have to be separated in calculating monitor units for complex situations, this approach has not been adopted within this report.

Volume-Scatter Ratio (V)

The volume-scatter ratio $V(z,s)$ is the ratio of the absorbed dose at depth z for a field size s at that depth in full-scatter conditions, to the dose in a mini-phantom at the same depth, same collimator setting and same source-detector distance. It therefore represents the fraction of additional scatter when measurements are performed in a large phantom rather than in a mini-phantom. It is assumed that V is independent of f , i.e. the divergence of the beam has a negligible effect on the value of V . A small influence of f on the tissue-phantom ratio has, however, been demonstrated (Björngård et al. 1996).

Scatter-Primary Ratio (S)

The scatter-primary ratio $S(z,c)$ is the ratio of the scatter contribution of absorbed dose and the primary contribution of absorbed dose, at depth z in a phantom for a field size c at that depth. A similarly defined quantity $S_0(z)$ is the scatter-primary ratio for a mini-phantom. $S_0(z)$ is independent of c .

It should be noted that scatter-primary ratio and volume-scatter ratio are related:

$$V(z,c) = \frac{[1 + S(z,c)]}{[1 + S_0(z)]} \quad (4.2)$$

The scatter contribution to absorbed dose can be determined by subtraction of the primary contribution from the total absorbed dose. The primary contribution can be determined by Monte Carlo calculations, or can be estimated by extrapolating the

variation of dose with field size at depth z to zero field size (Björngard and Vadash, 1995).

4.5.2 Depth of Dose Maximum.

At the surface of the irradiated material, kerma exceeds dose as electron equilibrium is not present. This situation changes with depth for two reasons. Firstly, the number of secondary electrons propagating within the irradiated material increases initially with depth. The rate of increase varies slightly with field size. Secondly, in the surface layers of the irradiated material are electrons present arising from electron contamination within the beam. The amount of this contamination depends upon distance, field size, and upon the presence of beam modifying devices (tray, compensator, wedge).

As a result, the depth at which maximum dose is achieved is not constant, but depends upon the operational parameters of the equipment. Because of this variation, the depth of dose maximum z_m is defined in this report as the greatest of these depths, usually for the smallest field size and the largest source-phantom distance used in practice.

At high beam energies, electron contamination occurs at depths considerably greater than the depth of dose maximum. The variation in this contamination with change in operational parameters may be considerable, producing distortions in quantities when they are normalised at this depth (percentage depth dose, tissue maximum ratio, tissue air ratio).

To overcome problems associated with uncertainties in dose determination at z_m , the recommendations within this report are that all quantities are normalized to a reference depth of 10 cm. The equations derived are, however, generic, and apply equally well at any large reference depth.

4.6 DOSE CALCULATION

Dose calculation is the process within treatment planning by which the magnitude and distribution of absorbed dose is calculated. It was outlined in section 4.2 as to why separate modelling of primary and scatter components may be desirable. This and subsequent sections described many of the features which need to be taken into account.

Calculation of variation in primary contribution to the dose is relatively straightforward in situations where electron equilibrium exists (although absolute

determination of the primary component is not so easy). Determination of scatter is more difficult. Rigorous calculation involves the summation of scatter from each volume element of the irradiated volume - a complex and time consuming process. In general, simplifications are adopted to facilitate computation.

4.6.1 Scatter Integration.

The scatter-primary ratio $S(z, c)$ is a function of depth, field size and field shape. It is generally assumed to be invariant with source-phantom distance.

Scatter integration is the process by which scatter from the various volume elements is summated. In situations where the primary contribution is constant across the beam, the scatter integration needs to account only for geometrical effects. The scatter to a point at depth z is given by:

$$D_s(z) = \frac{D_p(z)}{2\pi} \cdot \int S(r, \theta) d\theta \quad (4.3)$$

where $D_p(z)$ is the primary component, $S(r, \theta)$ is the scatter-primary ratio at a point of polar coordinates r and θ and the integral is taken around the perimeter of the field. Clarkson approximated this integral by a summation over a number of n sectors describing the field, and thereby determined an average value of S for those sectors (Clarkson, 1941):

$$D_s(z) = D_p(z) \cdot \bar{S} \quad \text{with} \quad \bar{S} = \frac{1}{2\pi} \sum_{i=1}^n S(r_i, \theta_i) \Delta\theta_i \quad (4.4)$$

where r_i is the radius of sector i , and $\Delta\theta_i$ is the angle subtended by the sector.

4.6.2 Equivalent Fields.

Circular or square fields equivalent to rectangular or irregular fields have been defined during many years as those fields leading to the same percentage depth-dose as the rectangular or irregular field under consideration, i.e. leading to the same value of S .

For any shape of field, there will be a circular field of radius R which has a value of S at depth z identical to that calculated from the above summation. This is termed the **equivalent radius** for the field. It is further assumed that such an equivalent radius is valid at any depth. An accurate approximation for points on the central axis of a square field of side s_e , is :

$$R = 0.561 s_e \quad (4.5)$$

For a rectangular field of size L by W , it may also be shown that the equivalent radius may be given approximately by:

$$R = 0.561 \cdot \frac{2LW}{L + W} \quad (4.6)$$

Combination of the above leads to the determination of an **equivalent square** for the rectangular field such that:

$$s_e = \frac{4LW}{2(L + W)} \quad (4.7)$$

$$\text{i.e. } s_e = 4 \cdot (\text{area/perimeter}).$$

This relation was originally developed by Sterling et al., (1964).

Detailed calculations of equivalent fields have been performed by Day and Aird (1983) for circular, square and rectangular fields.

These equations for equivalent fields cannot be used without verification for other parameters as for instance output ratios. The variation of output ratio is critically dependent upon head scatter which is associated with the details of the treatment head design. Other equations have been proposed as described in section 5.6.

4.6.3 Complex Situations.

The scatter integral presented in section 4.6.1 may be used to determine the scatter component for irregularly shaped fields. It may be generalised to situations where the primary component of the dose varies over the beam, or where the surface of the irradiated object is not normal to the central axis of the beam. Such situations require detailed integration over the cross-sectional area of the beam.

4.6.4 Pencil Beams.

An alternative approach is to consider the beam as being composed of many parallel pencil beams, each having a specific fluence.

Each pencil beam contributes to the dose at points both near to and far from the axis of the pencil. In the nearest region, the dose is deposited by secondary electrons produced by photons within the pencil. Energy is also transported by scattered photons which themselves produce secondary electrons at a distance from the pencil beam. Several generations of scattered photons may be involved. The function which expresses the dose as a function of depth and distance from the axis of the pencil is called a kernel. Kernels are calculated by Monte Carlo techniques and/or by fit to experimental results.

Modelling a photon beam requires a bundle of pencil beams of differing fluences and energies, and hence differing kernels. Computation involves adding the dose contributions from all pencil beams for each volume element of the irradiated volume, a process known as 'convolution' or 'superposition'. This type of calculation has the potential to handle situations involving lack of electron equilibrium.

4.6.5 Electron Transport Effects

(a) Transient electron equilibrium

As described in sections 4.3.2 and 4.3.3, the effects of electron transport need to be taken into account when there are gradients in the kerma distribution within the irradiated medium which result in absorbed dose not being equal to collision kerma. For transient electron equilibrium (see section 4.3.2) the gradient is small and the ratio of dose to collision kerma is about 1.02 and quite constant (Nilsson, 1985). Since dose and not kerma is measured, this perturbation is already taken into account in the beam data.

(b) Lateral electron equilibrium

Near the beam edges, dose will be less than kerma because electrons will be travelling towards the outside of the beam, with no corresponding electrons travelling inwards. This is taken into account when measured beam profiles extend beyond the penumbra region and are modelled with suitable functions. If the fields are very small, as may be the case in stereotactic applications, the dose on the central axis may also be affected by this process. This can be accounted for by using pencil-beam kernels, or by introducing additional measured factors.

5. MONITOR UNIT DETERMINATION IN OPEN BEAMS

5.1 INTRODUCTION

The square or rectangular beams defined by a conventional collimator with symmetrical jaws, are called open symmetrical beams when no beam modifier is present in the beam, such as wedge, compensator, tray or shielding blocks. Square and rectangular symmetrical beams obtained with asymmetrical collimators or multileaf collimators can be considered as open symmetrical beams in the absence of a beam modifier.

To calculate the number of monitor units U required to deliver the prescribed dose D_T to the patient, the physicist must always refer to what is generally called the monitor calibration, that is the dose per monitor unit (or per unit of time for ^{60}Co treatment units), $\dot{D}_R$ (see chapter 2, first paragraph), measured under reference conditions as defined, for instance, in the Technical Report 277 (IAEA, 1987). Two different sets of reference conditions corresponding to the two main irradiation set-ups have to be considered : isocentric conditions and fixed source-skin distance. They differ in general in relation to the distance from the source to the point of measurement and to the plane in which the field size is specified.

The choice of a reference depth much greater than the depth of maximum dose for monitor calibration has been dictated by the necessity to decrease the uncertainty associated with measurements at a depth where electron contamination can be of unpredictable influence. The same constraints apply to the measurements of the output ratio or to the normalisation of the tissue-phantom ratio T or the percentage depth dose P .

This document concentrates on measurements at a reference depth of 10 cm, but is similar to the approach of other authors (e.g. Khan 1984). These authors select a fixed reference depth, generally the depth of maximum dose, for normalisation of T or P , and for measurement of the output ratio.

In the next sections the basic quantities to be used in the dose calculation will be considered and simple equations derived for the calculation of the number of monitor units.

5.2 GEOMETRIC CONDITIONS

In most radiotherapy departments, it is routine to treat some patients in isocentric conditions and others at fixed source-skin distance.

In isocentric conditions the point of measurement for basic quantities is the isocentre and therefore the reference distance f_R , the source to isocentre distance, is the distance from the source to the point of measurement. In addition, the depth of the point of measurement z_R is specified as 10 cm, and the collimator setting c_R as 10 cm x 10 cm at the isocentre.

When a fixed source-skin distance is used, this distance f'_R is used as the reference distance. It is usually, but not necessarily, the source-axis distance for isocentric units (if so $f'_R = f_R$). The reference field size c_R is defined at the source-skin distance f'_R . Therefore the field size at the point of measurement, i.e. at the distance $f = f'_R + z_R$ from the source is equal to $c_R \times f / f'_R$.

There are several possible approaches for the determination of basic quantities in reference conditions:

a) We strongly recommend to refer to a unique set of reference measurements and to choose isocentric conditions as reference conditions. All the necessary quantities for fixed source-skin distance will be calculated from this unique set of reference data. For example, the dose per monitor unit, $\dot{D}'_R$ in reference conditions at a fixed source-skin distance f'_R . $\dot{D}'_R = \dot{D}(z_R, c_R, f'_R)$ (see section 5.5.2) is a derived quantity calculated from the dose per monitor unit $\dot{D}_R$ in reference isocentric conditions. The calculation of $\dot{D}'_R$ from $\dot{D}_R$ is not straightforward and does not depend only on the inverse-square law. The method to be used to generate information from a unique set of reference data, is described in Appendix A1 and Chapter 7.

We are aware that this is possibly a great change from the generally applied measurement conditions and may present difficulties. For this reason we propose other possibilities.

b) It is possible to use two different sets of reference measurements, including two different reference doses, for the two groups of patients. However, when more than 2 different geometric conditions are used on the same treatment unit, for example if more than one source-skin distance is used, the adoption of the second approach would lead to a multiplicity of reference data, increasing the possibility of incorrect selection of specific data.

c) For special techniques, such as Total Body Irradiation, Total Skin Irradiation and Stereotactic Radiotherapy, it is recommended to use different sets of reference measurements, including different reference doses, for the various groups of patients.

5.3 OUTPUT RATIO

As explained in chapter 3 the fluence of primary photons incident on the phantom is the sum of the photons originating in the target and of the head scattered photons. Consequently this fluence varies widely with the collimator setting. This variation is depicted by the **output ratio** which is the ratio of two doses measured at the same depth in the phantom and at the same distance from the target, for two different collimator settings. Two approaches are considered here. In the first one, the **output ratio** O_R is the ratio of two doses measured in a large water phantom and includes in addition the phantom scattered photons. In the second approach, the **output ratio** O_θ is the ratio of two doses measured in a mini-phantom in order to reduce the influence of the phantom scattered photons as far as possible.

5.3.1 Measurement in Full-Scatter Conditions

The output ratio O_R is measured in full-scatter conditions, in a large water phantom as:

$$O_R(c) = \frac{D(z_R, c)}{D(z_R, c_R)} \quad (5.1)$$

where z_R is the reference depth and c_R the reference collimator setting (figure 5-1).

The quantity O_R is easy to measure and is not sensitive to a possible electron contamination provided that z_R is sufficiently large. However it is a complex quantity since it accounts for both the head scatter variation and the phantom scatter variation between the field size of interest c , and the reference field size c_R . As treatment machines have each pair of collimating jaws at different distances from the source, the amount of head scatter within a rectangular beam, and therefore O_R , will depend upon the orientation of the rectangular field (see section 5.6). For the same reasons as mentioned for V , O_R is assumed to be independent of the distance from the point of interest to the source.

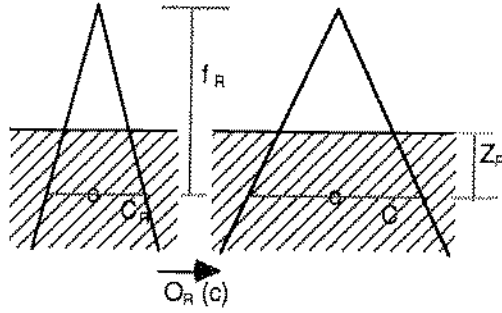


Figure 5-1 : Conditions of measurement of the output ratio $O_R(c)$ in isocentric conditions, as the ratio of the dose for a collimator setting c and a collimator setting c_R , in full scatter conditions.

5.3.2 Measurement in a Mini-Phantom

In order to separate as far as possible the contribution of the head scatter and the contribution of the phantom scatter to the dose, van Gasteren et al., (1991) have proposed to measure the dose D_0 at the reference depth z_R in a mini-phantom (figure 5-2).

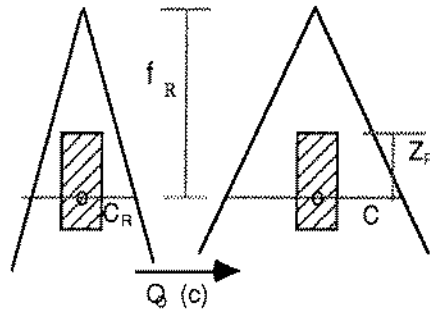


Figure 5-2 : Conditions of measurement of the output ratio $O_0(c)$ in isocentric conditions, as the ratio of the dose for a collimator setting c and a collimator setting c_R , in a mini-phantom.

The output ratio O_0 measured in a mini-phantom is :

$$O_0(c) = \frac{D_0(z_R, c)}{D_0(z_R, c_R)} \quad (5.2)$$

This ratio has been shown to be independent of the choice of z_R in the mini-phantom provided z_R is large enough to eliminate the contribution of electron contamination. The width of the mini-phantom is not critical, provided it is smaller

than the field size achieved at the reference distance by the smallest collimator setting used in clinical practice, and large enough to assure almost completely lateral electron equilibrium. A width of 4 cm seems to be realistic for x-rays up to 25 MV, since the diameter of the water volume necessary to achieve quasi-lateral electron equilibrium increases slowly with photon energy (Ahnesjö et al., 1987).

The ratio O_0 can be considered to be equal to the head scatter factor (Khan, 1984) since the contribution of the scattering by the mini-phantom is constant and small compared to the primary radiation, whatever the collimator setting is, and appears in both numerator and denominator. O_0 does not depend strongly on measurement distance, being similar for all distances usually encountered in clinical practice. It is, however, generally measured at the reference distance. As will be discussed in section 5.6, both O_0 and O_R depend upon the orientation of the collimator for rectangular beams.

5.4 ISOCENTRIC CONDITIONS

5.4.1 General Conditions

5.4.1.1 Reference conditions

As stated in section 5.2, under isocentric conditions the point of measurement for basic quantities is the isocentre, the reference distance f_R is the distance from the source to the isocentre and c_R is the collimator setting at the isocentre.

5.4.1.2 Tissue-phantom ratio

The tissue-phantom ratio $T(z,s)$ (see section 4.5.1) 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 :

$$T(z,s) = \frac{D(z,s,f)}{D(z_R,s,f)} \quad (5.3)$$

Under clinical conditions, the tissue-phantom ratio does not depend on f , as has been discussed for the volume-scatter ratio, but is a function of the field size s at f . When the distance f is the reference distance f_R , $T(z,s)$ can be written as $T(z,c)$.

5.4.2 Dose Calculation in Full-Scatter Conditions

In full-scatter isocentric conditions, the dose is measured at a constant reference distance f_R from the source, equal to the distance between the source and

the axis of rotation of the treatment unit. The field size at the depth of measurement is the field size indicated by the collimator setting c . In these geometrical conditions, using the output ratio measured in reference conditions $O_R(c)$, the dose $D(z,c)$ at a depth z , collimator setting c and a source-detector distance f_R is calculated by the following equation :

$$D(z,c) = \dot{D}_R \cdot U \cdot O_R(c) \cdot T(z,c) \quad (5.4)$$

where $\dot{D}_R = \dot{D}(z_R, c_R, f_R)$ is the dose per monitor unit or per timer unit in reference conditions and U is the number of monitor units or time units required to deliver the dose D at depth z for the collimator setting c (figure 5-3).

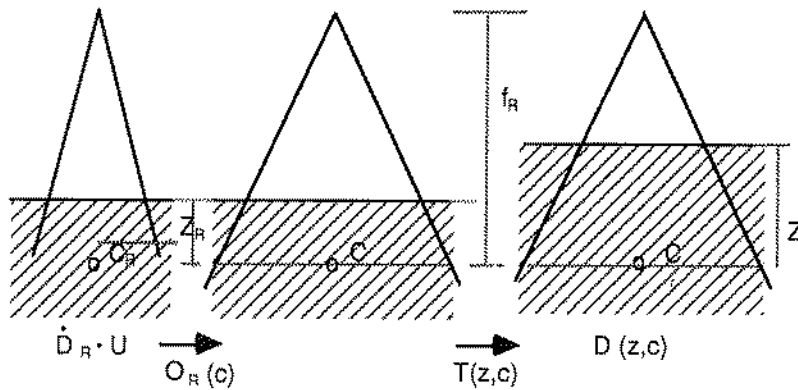


Figure 5-3 : Conditions of measurement for the derivation of the dose $D(z,c)$, in isocentric conditions, from the dose per monitor unit or per unit of time $\dot{D}_R$ in reference conditions, and the output ratio $O_R(c)$ in full scatter conditions.

5.4.3 Dose Calculation when a Mini-Phantom is used

5.4.3.1 Derivation of basic quantities from measurement with a mini-phantom

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

The ratio V of the dose $D(z,s)$ at a depth z for a field size s at that depth in a full-scatter water phantom, and the dose $D_0(z,s)$ in a mini-phantom for the same depth, same collimator setting and same source-detector distance f , represents the contribution to the dose of the photons scattered by the additional phantom material when the scattering volume is increased from the mini-phantom to the large phantom:

$$V(z,s) = \frac{D(z,s,f)}{D_0(z,s,f)} \quad (5.5)$$

The volume scatter ratio $V(z,s)$ depends upon the field size s at the depth of measurement, but is not, in a first approximation, a function of the source-detector distance, provided that the two doses are measured at the same distance. However, as the beam size s varies with the distance for a given collimator setting c , $V(z,s)=V(z,c)$ only when the distance f to the source is the reference distance f_R .

The quantity $V(z_R,c) = D(z_R,c)/D_0(z_R,c)$, the ratio of the doses at the reference depth in a large phantom and in the mini-phantom, is a function of the collimator setting. This volume scatter ratio at the reference depth is easy to measure (see figures 5-1 and 5-2).

When f is equal to f_R , the ratio of $V(z_R,c)$ and $V(z_R,c_R)$ is proportional to the ratio of the output ratios O_R and O_0 :

$$\frac{V(z_R,c)}{V(z_R,c_R)} = \frac{O_R(c)}{O_0(c)} \quad (5.6)$$

This ratio is identical to the phantom scatter correction factor $S_p(z_R,c)$ as defined by several groups, e.g. Khan (1984) and van Gasteren et al., (1991).

The scatter-primary ratio $S(z,c)$ for the beam size c (see 4.2) can be related to $V(z,c)$ and $S_0(z)$ where $S_0(z)$ is the scatter-primary ratio for the mini-phantom. The ratio $S_0(z)$ increases with depth, decreases with increasing energy and is independent of field size. It cannot easily be measured but can be approximated by theoretical calculation.

The tissue-phantom ratio T and the volume scatter ratio V are related by:

$$T(z,s) = \frac{V(z,s)}{V(z_R,s)} \cdot \frac{D_0(z,s)}{D_0(z_R,s)} \quad (5.7)$$

Both T and V depend upon depth z and field size s at the depth of measurement, but not on the distance from the source.

For open beams, and when $f = f_R$, equation 5.7 becomes :

$$T(z,c) = \frac{V(z,c)}{V(z_R,c)} \cdot \frac{D_0(z,c)}{D_0(z_R,c)} \quad (5.8)$$

where the second term on the right hand represents the beam attenuation in the mini-phantom $(T_0)_{z/z_R}$ by a thickness $z-z_R$ of material. This attenuation in the mini-phantom is independent of s (or c) and f when z is larger than z_R (van Gasteren et al., 1991).

5.4.3.2 Dose calculation when a mini-phantom is used

Using relation 5.6 between V and O , equation 5.4 can be rewritten as :

$$D(z, c) = \dot{D}_R \cdot U \cdot O_0(c) \cdot \frac{V(z_R, c)}{V(z_R, c_R)} \cdot T(z, c) \quad (5.9)$$

where the output ratio $O_0(c)$ accounts for variations in head scatter, and the last two terms account for attenuation and scattering variations in the large water phantom (figure 5-4). The ratio of volume scatter terms V , represents the contribution of the phantom scatter at the reference depth when the beam size varies from c_R to c . This ratio is larger or smaller than 1, depending on whether c is larger or smaller than c_R . Theoretical and experimental determinations show that this ratio depends on $T_{2w,10}$ but is not very sensitive to the type of accelerator or to the radial energy variation (Storchi and van Gasteren 1996, S  therberg et al., 1996).

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

5.5 FIXED SOURCE-SKIN DISTANCE

When most of the treatments are performed at a fixed source-skin distance, the use of tissue-phantom ratio is not straightforward and other quantities are often preferred.

5.5.1 General Considerations

5.5.1.1 Reference conditions

When a fixed source-skin distance is used, the total distance f is the reference distance f'_R plus the depth of the point of measurement in reference conditions:

$$f = f'_R + z_R \quad (5.10)$$

At the point of measurement of the output ratio, the distance from the source is f and therefore the field size is $c \cdot f / f'_R$.

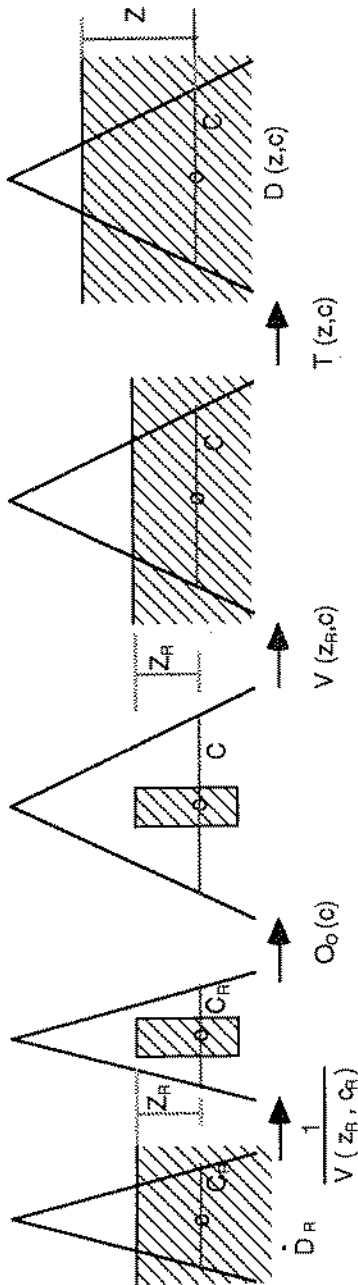


Figure 5-4 : Conditions of measurement for the derivation of the dose $D(z, c)$, in isocentric conditions, from the dose per monitor unit or per unit of time $\dot{D}_R$ in reference conditions, and the output ratio $O_R(c)$ in a mini-phantom.

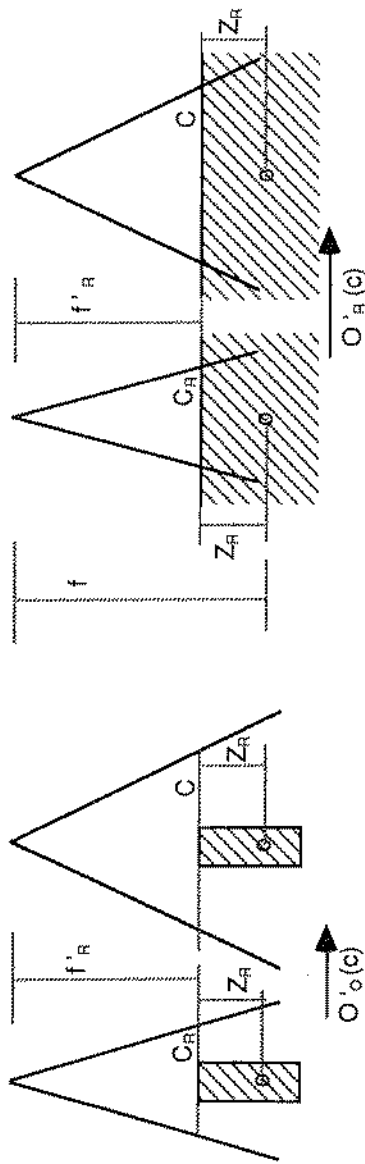


Figure 5-5 : Conditions of measurement of the output ratios $O'_0(c)$ and $O'_R(c)$, at a fixed source-skin distance, as the ratios of the dose for a collimator setting c and a collimator setting c_R , in a mini-phantom and in full scatter conditions, respectively.

To avoid any confusion, when O_0 and O_R are measured at a distance $f = f'_R + z_R$ between the source and the point of measurement, they are denoted O'_0 and O'_R . They refer to a field size c at the surface and not at depth z_R . For similar reasons, at fixed source-skin distance, doses will be labelled D' (figure 5-5).

5.5.1.2 Percentage depth dose

The conventional definition of the percentage depth dose $P(z, s, f')$ (see section 4.5.1) is the ratio of the dose $D'(z, s, f')$ at depth z , and the maximum dose $D'(z_m, s, f')$ along the beam axis, expressed as a percentage.

$$P(z, s, f') = \frac{D'(z, s, f')}{D'(z_m, s, f')} \cdot 100 \quad (5.11)$$

where f' is the fixed source-skin distance.

As it is recommended in this report to refer the basic quantities always to the reference depth z_R and not to the depth of maximum dose z_m , it is recommended to replace the conventional percentage depth dose P by the reference percentage depth dose P_R :

$$P_R(z, s, f') = \frac{D'(z, s, f')}{D'(z_R, s, f')} \cdot 100 \quad (5.12)$$

If the source-skin distance is equal to the source-isocentre distance the field size s is expressed by the collimator setting c and $P(z, s, f')$ can be written $P(z, c, f')$.

5.5.2 Dose Calculation in Full-Scatter Conditions

Under the full-scatter conditions of a large phantom at a fixed source-skin distance equal to the source-isocentre distance, the dose can be calculated as :

$$D'(z, c, f'_R) = D'_R \cdot U \cdot O'_R(c) \cdot P_R(z, c, f'_R) / 100 \quad (5.13)$$

where $D'_R = D'(z_R, c_R, f'_R)$ is the dose per monitor unit in reference conditions and U is the number of monitor units (figure 5-6).

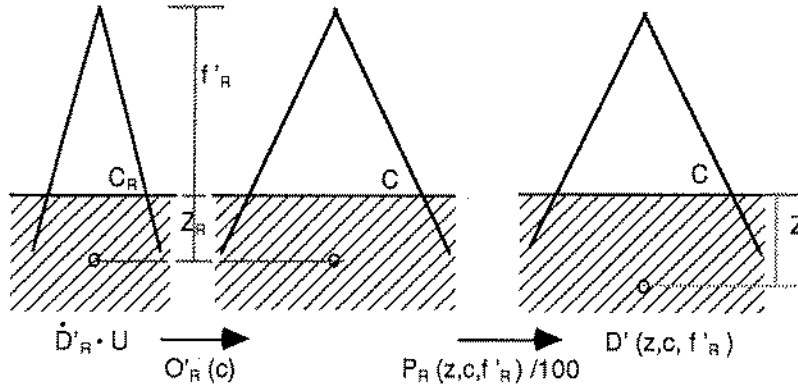


Figure 5-6 : Conditions of measurement for the derivation of the dose $D'(z, c, f'_R)$, at a fixed source-skin distance, from the dose per monitor unit or per unit of time $\dot{D}'_R$ in reference conditions, and the output ratio $O'_R(c)$ in full scatter conditions.

5.5.3 Dose Calculation when a Mini-Phantom is used

In equation 5.13 one can replace O'_R by O'_0 :

$$O'_R(c) = O'_0(c) \cdot \frac{V(z_R, c \times f / f'_R)}{V(z_R, c_R \times f / f'_R)} \quad (5.14)$$

and then calculate the dose with a separation of the various physical quantities :

$$D'(z, c, f'_R) = \dot{D}'_R \cdot U \cdot O'_0(c) \cdot \frac{V(z_R, c \times f / f'_R)}{V(z_R, c_R \times f / f'_R)} \cdot P_R(z, c, f'_R) / 100 \quad (5.15)$$

This separation into head scatter and phantom scatter terms is especially useful when shielding blocks are used (figure 5-7).

5.6 RECTANGULAR FIELDS

Equations (5.1) to (5.15) are strictly valid only for square fields. The case of irregular fields defined by additional blocks will be considered extensively in section 6.3 but most of the prescribed fields are not square but rectangular. The use of the previous equations for rectangular fields is not straightforward because the determination of an equivalent square for a rectangular field differs for head scatter and for phantom scatter.

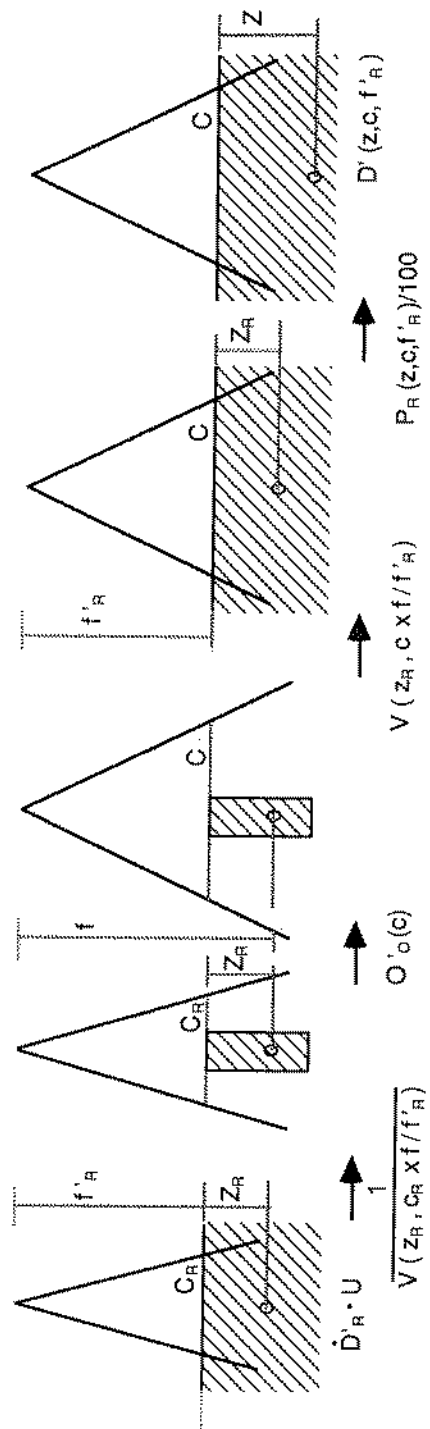


Figure 5-7 : Conditions of measurement for the derivation of the dose $D'(z, c, f'_R)$, at a fixed source-skin distance, from the dose per monitor unit or per unit of time $\dot{D}'_R$ in reference conditions, and the output ratio $O'_o(c)$ in a mini-phantom.

The conventional approach (see 4.6.2) is that the scatter-primary ratio S of a given rectangular field for points on the central axis, can be calculated with the conventional equivalent square s_e as given by equation (4.7). s_e should therefore be used in all quantities where phantom scatter variation has to be considered, such as V , T or P .

Because of the collimator design, the head scatter and consequently O_0 , depends upon the orientation of the collimator for rectangular beams. This effect is generally called the collimator exchange effect (e.g. Kase and Svensson 1986, Pattersson and Shragge 1981). The collimator exchange effect is mainly caused by a difference in scattered radiation that can reach the point of measurement for the same collimator setting of the upper and lower jaws. For some types of accelerators the difference in monitor backscatter for upper and lower jaws also contributes to the collimator exchange effect (Lam et al., 1996).

If accurate values of O_0 are desired, different combinations of collimator width W and length L have to be considered. Various methods of parametrisation of O_0 have been proposed, based upon fitting a specific function to the measured data (Chen 1990, Szymczyk et al. 1991, Vadash and Bjärngard 1993, Yu et al. 1995). In another approach, physical models have been developed to describe the photon energy fluence reaching the point of calculation (Ahnesjö et al. 1992, Dunscombe and Nieminen 1992, Yu and Sloboda 1996, Lam et al. 1996). A review of the various parametrisation methods has recently been given by Jager et al (1996).

Most methods of parametrisation of O_0 apply the equivalent field size method. Because equation 4.7 is symmetrical in L and W more sophisticated equations have been suggested. As an example, Vadash and Bjärngard (1993) have suggested the following equation for a **collimator equivalent square** c_e to a rectangular collimator setting (L, W) , where L and W are the openings of the outer and inner jaws respectively:

$$c_e = (A+1)LW/(AL+W) \quad (5.16)$$

different values of A have been suggested, depending on the collimator design and the beam energy. For $A=1$, the Sterling et al. formula (Eq. 4.7) is obtained.

The collimator equivalent square c_e should be used in the quantities O_0 and O_R which account for the head scatter variation. Because O_R accounts for both the head scatter variation and the phantom scatter variation, the collimator equivalent square c_e for O_R differs from the value determined for O_0 . Therefore the parameters

used should be carefully determined in the measurement conditions of either O_0 or O_R .

Equations (5.4), (5.9), (5.13) and (5.15) can be transformed into equations (5.17), (5.18), (5.19) and (5.20), respectively, for rectangular fields, where c in the first part of the equations, stands for the collimator setting L, W :

$$D(z, c) = \dot{D}_R \cdot U \cdot O_R(c_e) \cdot T(z, s_e) \quad (5.17)$$

$$D(z, c) = \dot{D}_R \cdot U \cdot O_0(c_e) \cdot \frac{V(z_R, s_e)}{V(z_R, c_R)} \cdot T(z, s_e) \quad (5.18)$$

$$D'(z, c, f'_R) = \dot{D}'_R \cdot U \cdot O'_R(c_e) \cdot P_R(z, s_e, f'_R) / 100 \quad (5.19)$$

$$D'(z, c, f'_R) = \dot{D}'_R \cdot U \cdot O'_0(c_e) \cdot \frac{V(z_R, s_e \times f / f'_R)}{V(z_R, c_R \times f / f'_R)} \cdot P_R(z, s_e, f'_R) / 100 \quad (5.20)$$

5.7 CALCULATION BASED ON THE ENERGY FLUENCE

The method to calculate the number of monitor units required to deliver the prescribed dose is somewhat different with algorithms that model the radiation beam from its incident energy fluence Ψ and arrive at the dose by superposition or convolution of kernels or by direct Monte-Carlo simulation. However, the principle is the same as with the traditional algorithms in that the starting-point is the calibration of the beam monitor in the reference geometry, i.e. from Eq. 5.4 :

$$D(z_R, c_R) \equiv \dot{D}_R \cdot U \quad (5.21)$$

The difference is the methodology to arrive at the dose $D(z, c)$ at the point of interest. This methodology will not be described in detail, only the architecture of the relationship between Ψ , U , $D(z, c)$ and $D(z_R, c_R)$.

With the identity of Eq. (5.21) we can make the trivial but important statement

$$D(z, c) = \dot{D}_R \cdot U \cdot \frac{D(z, c)}{D(z_R, c_R)} \quad (5.22)$$

In traditional methods, the ratio $D(z, c)/D(z_R, c_R)$ represents experimental data, for instance as in Eq. 5.9,

$$D(z, c) = \dot{D}_R \cdot U \cdot O_0(c) \cdot \frac{V(z_R, c)}{V(z_R, c_R)} \cdot T(z, c) \quad (5.23)$$

This expression may be expanded further using measured ratios to correct for deviations from the idealised geometrical conditions in which the tissue-phantom ratio and the output ratio were measured.

In fluence-based calculations, a similar approach is used. The identity

$$\frac{D(z, c)}{D(z_R, c_R)} = \frac{D(z, c)}{\Psi(c_R)} \cdot \frac{\Psi(c_R)}{D(z_R, c_R)} \quad (5.24)$$

yields

$$D(z, c) = \dot{D}_R \cdot U \cdot \frac{\Psi(c_R)}{D(z_R, c_R)} \cdot \frac{D(z, c)}{\Psi(c_R)} \quad (5.25)$$

Again, further expansion using calculated or measured ratios to correct for additional processes is possible as long as the fundamental identity Eq.5.21 is retained. For instance, Eq.5.26 is analogous to Eq.5.23 :

$$D(z, c) = \dot{D}_R \cdot U \cdot \frac{\Psi(c)}{\Psi(c_R)} \cdot \frac{\Psi(c_R)}{D(z_R, c_R)} \cdot \frac{D(z, c)}{\Psi(c)} \quad (5.26)$$

Here, $\Psi(c)/\Psi(c_R)$ corresponds to the output factor in air and accounts for the same physical effects, although it is defined in terms of energy fluence, as $O_0(c)$. The ratio $D(z, c)/\Psi(c)$ corresponds to the product of $T(z, c)$ by $V(z_R, c)$ in Eq.5.23 but is normalised to the incident fluence for the same collimator setting rather than the dose at the reference depth.

The ratio $\Psi(c_R)/D(z_R, c_R)$ corresponds, but is not identical, to $1/V(z_R, c_R)$, because $\dot{D}_0(z_R, c_R)$ is not equal to $\Psi(c_R)$.

Both ratios in Eq.5.25 for instance are calculated, one for the reference geometry and one for the patient geometry. The product of these two factors is $D(z, c)/D(z_R, c_R)$ and must agree with measured data.

6. MONITOR UNIT DETERMINATION IN MODIFIED BEAMS

6.1. INTRODUCTION

The influence of beam modifiers, such as shielding blocks and wedge filters, on the dose, can be taken into account quantitatively in two ways.

The conventional method is to refer to dose measurements performed in a large phantom. This procedure takes into consideration the combined modifications in head and in phantom scatter due to the presence of beam modifiers in the beam.

The second method is to consider the changes in the head scatter independently from the modification in scattered radiation produced in the phantom by the accessories interposed in the beam. Practically, the separation of dose components coming from the head of the treatment unit and the phantom requires measurements performed in the mini-phantom described previously (paragraph 5.3.2) and in a large water phantom.

This method is particularly well adapted when dose calculations have to be performed for complex treatment conditions using modified beams or referring to a treatment distance different from the reference distance. In this report, both methods of dose calculation are presented for wedged and blocked fields. For each type of modified beam (wedges or blocks), equations are presented both for isocentric conditions (sections 6.2.2 and 6.3.2) and fixed source-skin distance (section 6.2.3 and 6.3.3). The equations presented in this section relate only to points located along the central axis. It is furthermore assumed that the formalisms for wedges and blocks can be applied independently ; i.e. if both blocks and wedges are used, a combination of both procedures is valid.

6.2. WEDGED BEAMS

6.2.1. General Concepts

Wedge filters are often used in clinical practice to modify the dose distribution in the patient. The modification of the dose distribution results in a slope of isodoses being turned through an angle related to the wedge angle. There is a corresponding decrease of the output of the treatment unit.

The calculation of the number of monitor units in a wedged photon beam is frequently based on the use of an unique correction factor. This factor, called wedge factor, is defined as the ratio of the absorbed doses in water at a point along the beam

axis, with and without the wedge under reference conditions, for the same number of monitor units (ICRU 1976).

Several authors (e.g. Bjärngård and Shackford, 1994, Georg et al., 1997, Heukelom et al., 1994a, McCullough et al., 1986, Niroomand-Rad et al., 1992, Palta et al., 1988) have shown however that the wedge factor varies with field size and depth in a phantom.

This variation of the wedge factor with field size at a given depth has been related to the filtration of the scattered radiation from the treatment head by the wedge filter, to the addition of photons scattered by the wedge and, to a lesser degree, to the changes in the amount of backscatter into the monitor of linear accelerators.

The variation of the wedge factor with depth in a phantom has been explained by beam hardening through the wedge filter which absorbs and/or scatters the low-energy photon beams. This effect leads to a change in the beam quality and consequently to a modification of the depth dose distribution. The magnitude of this modification depends upon the energy spectrum of the open beams and on the composition and thickness of the wedge.

The reference conditions of field size and depth in phantom used for the determination of the wedge factor are not well defined in the literature. For example, no recommendation is given in ICRU Report 24 (ICRU, 1976). Some authors specify a reference depth corresponding to the depth of dose maximum while others refer to a larger depth, (5 cm or 10 cm), the measurements being performed in a water or water-equivalent phantom of dimensions large enough to ensure full scatter.

As a first approximation, the calculation of the number of monitor units for wedged fields can be made by using the depth dose data (percentage depth dose or tissue-phantom ratios) of the unwedged field and an average wedge factor measured under reference conditions. This method does not take into account the variation of the wedge factor with depth and field size, which may significantly affect the result.

The influence of the wedge filter on the dose at a given point in water can be determined accurately either by using a wedge factor measured in a large water phantom at the reference depth for the reference field size in conjunction with other factors (e.g. *O*, *P*, *T*) measured specifically for the wedge under consideration, or by using a wedge factor which varies with field size and depth in conjunction with unwedged data for these other factors.

In a second method, the wedge factor is measured in the mini-phantom (as described in Chapter 5) under reference conditions and the depth dose is measured separately in a large water phantom for each wedged field.

The present chapter reports the two methods of dose calculation using the wedge factor determined with the mini-phantom or in a large water phantom. The application of the derived relationships to the calculation of the number of monitor units when a treatment is performed with a wedge filter present during part of the treatment, is discussed in detail in Appendix A.2.

6.2.2. Isocentric Conditions

The reference conditions (depth z_R , field size c_R and distance between the source and the point of interest f_R) are identical to that used for open beams (see chapter 5). The field size c_R is defined at f_R .

6.2.2.1. Dose calculation based on measurements with a mini-phantom

6.2.2.1.1. The wedge factor

$$k_{o,w}(c_R) = \frac{D_o(z_R, c_R, f_R, w)}{D_o(z_R, c_R, f_R)} \quad (6.1)$$

The **wedge factor** $k_{o,w}(c_R)$ under reference conditions of field size c_R is the ratio of the doses measured with the mini-phantom at the reference distance f_R , with and without the wedge filter in the beam, for the same number of monitor units (figure 6-1). It takes into account the modification of the head scatter produced across the wedge filter.

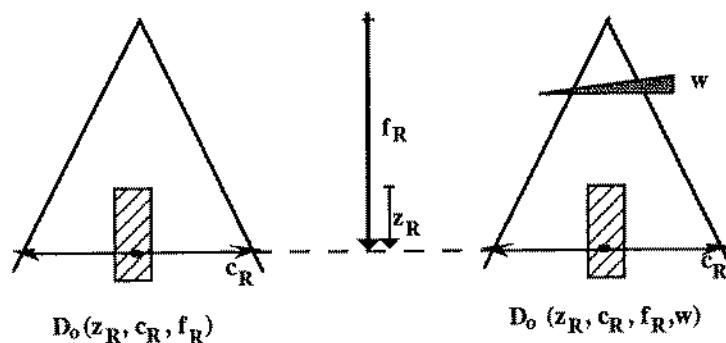


Figure 6-1. Diagram illustrating the definition of the wedge factor $k_{o,w}(c_R)$ measured under isocentric conditions with a mini-phantom.

6.2.2.1.2. Dose calculation

The dose under treatment conditions $D(z, c, f_R, w)$ in a wedged beam can be calculated from the dose per monitor unit under reference conditions $\dot{D}_R$ in an open beam according to the following relation (see figure 6-2).

$$D(z, c, f_R, w) = \dot{D}_R \cdot U \cdot O_\theta(c, w) \cdot \frac{V(z_R, c, w)}{V(z_R, c_R)} \cdot k_{o,w}(c_R) \cdot T(z, c, w) \quad (6.2)$$

- $O_\theta(c, w)$ refers to the output ratio for field size c and with the wedge filter in the beam measured with the mini-phantom,
- $\frac{V(z_R, c, w)}{V(z_R, c_R)}$ is the ratio of the volume scatter ratios for the field size c with the wedge filter and for the field size c_R without the wedge filter.
- $k_{o,w}(c_R)$ is the wedge factor measured in the reference field (c_R) and distance (f_R) with the mini-phantom,
- $T(z, c, w)$ is the tissue-phantom ratio at depth z , for field size c with the wedge filter in the beam.

6.2.2.2. Dose calculation based on measurements in a large water phantom

6.2.2.2.1. The wedge factor

a) Under reference conditions

$$k_w(z_R, c_R) = \frac{D(z_R, c_R, f_R, w)}{D(z_R, c_R, f_R)} \quad (6.3)$$

The wedge factor $k_w(z_R, c_R)$ in reference conditions (depth in water z_R and field size c_R) is the ratio of the doses in water measured at the reference distance f_R from the source with and without the wedge in the beam, for the same number of monitor units (figure 6-3).

b) Under treatment conditions

The wedge factor $k_w(z, c)$ under treatment conditions at depth z and field size c represents the ratio of the doses in water measured under these conditions at the reference distance f_R with and without the wedge filter in the beam, for the same number of monitor units.

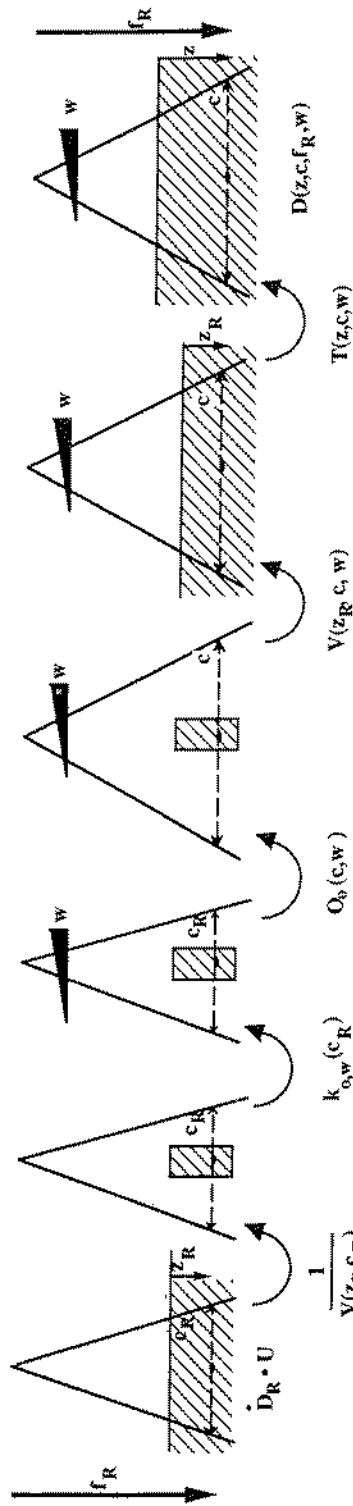


Figure 6-2. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a wedged beam from the dose $\dot{D}_R$ per monitor unit under reference conditions in an open beam when the treatment is performed in isocentric conditions at the reference distance f_R and when referring to measurements with a mini-phantom.

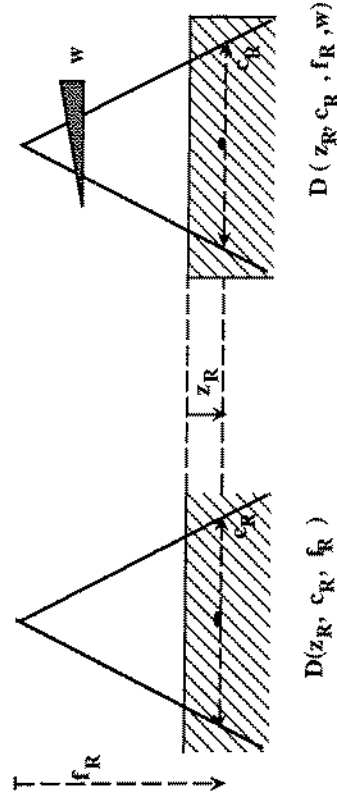


Figure 6-3. Diagram illustrating the definition of the wedge factor $k_w(z_R, c_R)$ measured in isocentric conditions in a large water phantom.

$k_w(z, c)$ can be measured directly or derived from the wedge factor $k_w(z_R, c_R)$ determined under reference conditions by taking into account the relative variation of the wedge factor with depth and field size. The expression of $k_w(z, c)$ is :

$$k_w(z, c) = k_w(z_R, c_R) \cdot a_w(z_R, c) \cdot b_w(z, c) \quad (6.4)$$

where

$a_w(z_R, c)$ is a relative factor which takes into account the field size dependence of the wedge factor at the reference depth. It corresponds to the ratio of the output factor in water for the field size c for the wedged beam and for the corresponding open beam :

$$a_w(z_R, c) = \frac{O_R(c, w)}{O_R(c)} \quad (6.5)$$

and $b_w(z, c)$ is a relative factor which takes into account the depth dependence of the wedge factor for the field size under consideration. It is equal to the ratio of the tissue-phantom ratios at depth z for the field size c for the wedged beam and open beam, respectively:

$$b_w(z, c) = \frac{T(z, c, w)}{T(z, c)} \quad (6.6)$$

In isocentric conditions, $k_w(z_R, c_R)$, $a_w(z_R, c)$ and $b_w(z, c)$ are measured at a distance between the source and the point of interest equal to f_R .

By combining relations (6.4, 6.5, 6.6), the expression of $k_w(z, c)$ is :

$$k_w(z, c) = k_w(z_R, c_R) \cdot \frac{O_R(c, w)}{O_R(c)} \cdot \frac{T(z, c, w)}{T(z, c)} \quad (6.7)$$

6.2.2.2.2. Dose calculation

On the basis of the previous considerations, the absorbed dose under the presence of the wedge filter can be calculated from the absorbed dose per monitor unit $\dot{D}_R$ determined under reference conditions of the open beam

- either by using the "reference wedge factor" $k_w(z_R, c_R)$ together with depth and field size factors which relate to the to the wedged beam :

$$D(z, c, f_R, w) = \dot{D}_R \cdot U \cdot O_R(c, w) \cdot k_w(z_R, c_R) \cdot T(z, c, w) \quad (6.8)$$

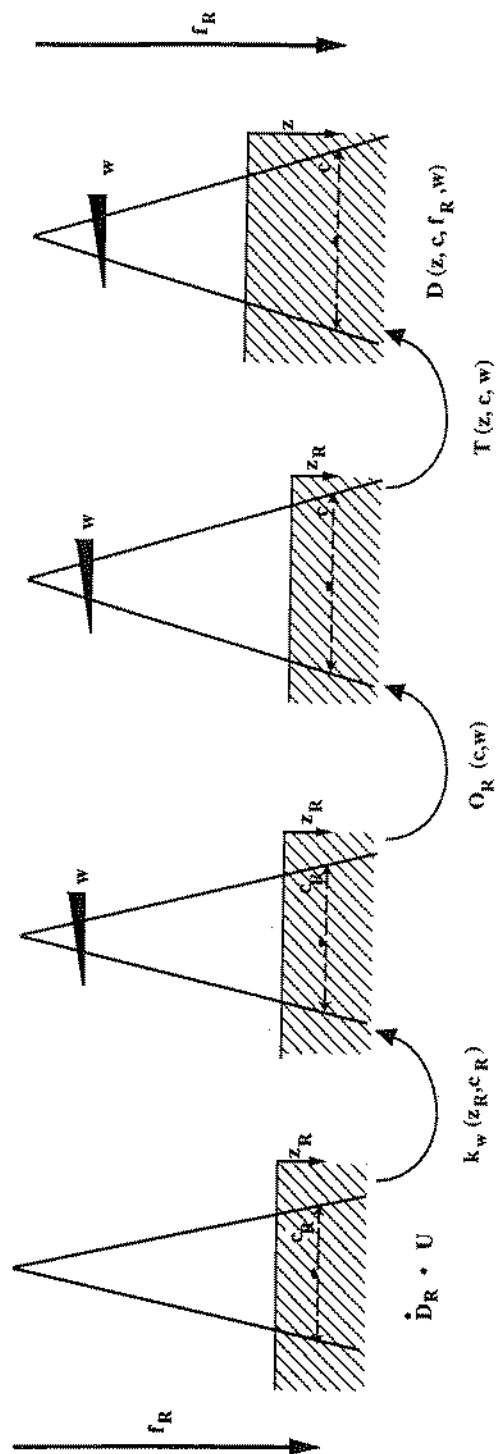


Figure 6-4. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a wedged beam from the dose per monitor unit under reference conditions in an open beam $\dot{D}_R$ when the treatment is performed isocentric conditions at the reference distance f_R and when referring to measurements in a large water phantom.

Figure 6-4 describes the different steps used to derive D from $\dot{D}_R$.

- or, by using the wedge factor in treatment conditions $k_w(z, c)$ together with depth and field size factors which relate to the open beam :

$$D(z, c, f_R, w) = \dot{D}_R \cdot U \cdot O_R(c) \cdot k_w(z, c) \cdot T(z, c) \quad (6.9)$$

The wedge factor $k_w(z, c)$ can be derived from $k_w(z_R, c_R)$ according to relation 6.4 or 6.7. This method requires the experimental determination of the relative variation of the wedge factor with depth and field size.

The two equations (relations 6.8 and 6.9) are equivalent and can be derived easily from each other by use of equation 6.7.

Whichever method is chosen, careful attention must be paid to the parameters needed for calculation : some relate to the wedged beam, others to the open beam.

The algorithm of any computer programme must be checked to identify which (if any) of the factors needed for the calculation of monitor units are integrated into the software, and which factors need to be applied externally.

6.2.2.3. Relation between the wedge factors determined in a large water phantom and with a mini-phantom.

The wedge factor at the reference depth z_R and for the reference field size c_R in a large water phantom is related to the wedge factor measured in a mini-phantom under the same circumstances according to :

$$k_w(z_R, c_R) = k_{o,w}(c_R) \cdot \frac{V(z_R, c_R, w)}{V(z_R, c_R)} \quad (6.10)$$

$k_{o,w}(c_R)$ takes into account the change in the head scatter by the wedge filter whereas $k_w(z_R, c_R)$ includes in addition the change in the phantom scatter (Bar-Deroma and Bjärngård, 1994, Heukelom et al., 1994b).

6.2.3. Fixed Source-Skin Distance

For the treatment performed with the fixed source-skin distance technique, the reference distance and the field size are defined at the skin of the patient and are f'_R and c_R , respectively. The reference depth is z_R .

The distance f'_R is assumed to be equal to the source-rotation axis of the treatment unit.

6.2.3.1. Dose calculation based on measurements with a mini-phantom

6.2.3.1.1. The wedge factor

$$k'_{o,w}(c_R) = \frac{D_o(z_R, c_R, f'_R, w)}{D_o(z_R, c_R, f'_R)} \quad (6.11)$$

The wedge factor $k'_{o,w}(c_R)$ in reference conditions of field size c_R is the ratio of the doses measured with the mini-phantom at the reference distance f'_R , with and without the wedge filter in the beam (figure 6-5), for the same number of monitor units.

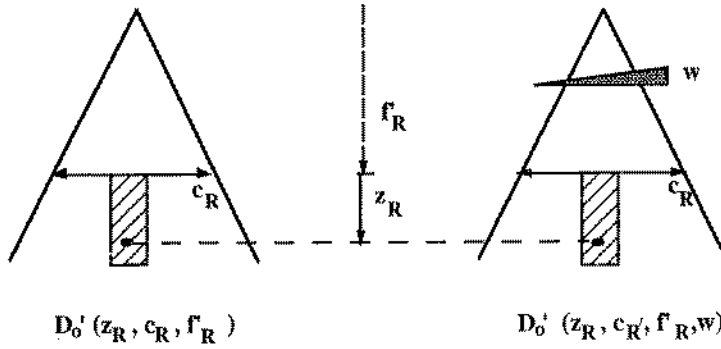


Figure 6-5. Diagram illustrating the definition of the wedge factor $k'_{o,w}(c_R, z_R)$ measured at the reference source-skin distance f'_R with a mini-phantom.

6.2.3.1.2. Dose calculation

The relation between $D'(z, c, f'_R, w)$ and $\dot{D}'_R$ is, according to figure 6-6 :

$$D'(z, c, f'_R, w) = \dot{D}'_R \cdot U \cdot O'_0(c, w) \cdot \frac{V(z_R, c', w)}{V(z_R, c_R)} \cdot k'_{o,w}(c_R) \cdot \frac{P_R(z, c, f'_R, w)}{100} \quad (6.12)$$

where :

$$c' = c \cdot \frac{f'_R + z_R}{f'_R}$$

$$c'_R = c_R \cdot \frac{f'_R + z_R}{f'_R}$$

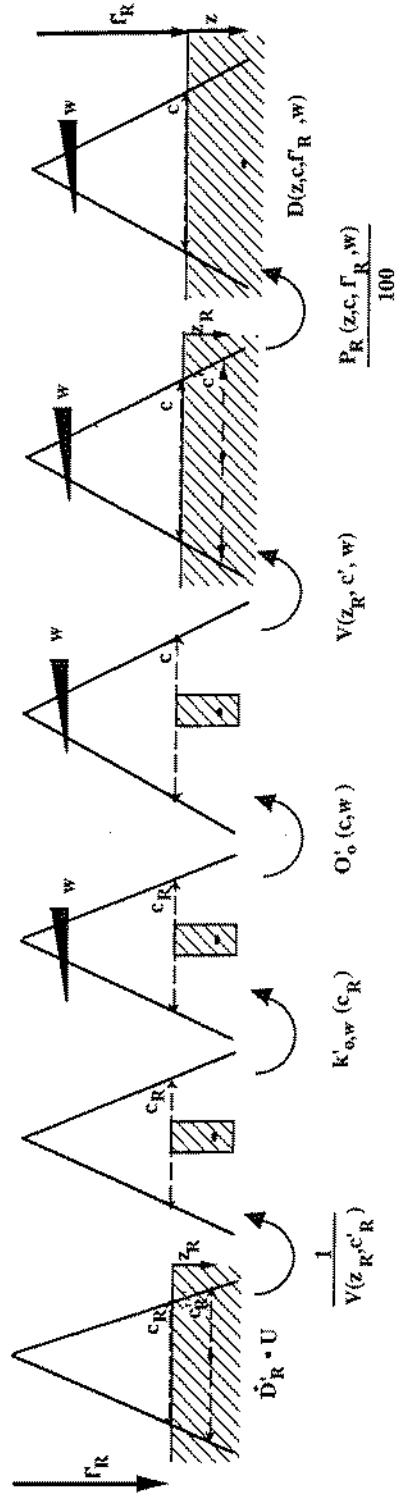


Figure 6-6. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a wedged beam from the dose per monitor unit under reference conditions in an open beam $\dot{D}_R$ when the treatment is performed at the source-skin distance f_R and when referring to measurements with a mini-phantom.

6.2.3.2. Dose calculation based on measurements in a large water phantom

6.2.3.2.1. The wedge factor

a) Under reference conditions

$$k'_w(z_R, c_R) = \frac{D(z_R, c_R, f'_R, w)}{D(z_R, c_R, f'_R)} \quad (6.13)$$

The wedge factor $k'_w(z_R, c_R)$ under reference conditions (depth z_R and field size c_R) is the ratio of the doses measured at the reference source-skin distance f'_R , with and without the wedge in the beam, for the same number of monitor units (figure 6-7).

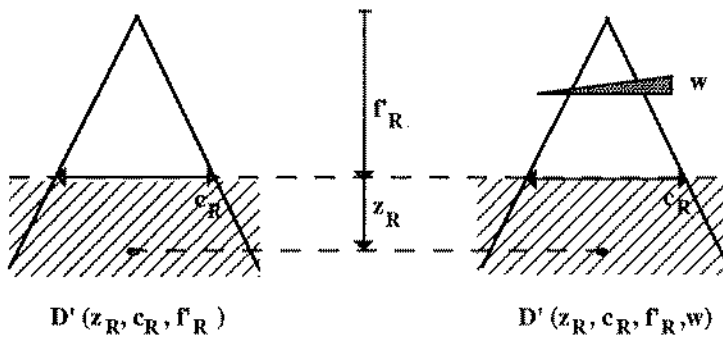


Figure 6-7. Diagram illustrating the definition of the wedge factor $k'_w(c_R, z_R)$ measured at the reference source skin-distance f'_R in a large water phantom.

b) Under treatment conditions

The wedge factor $k'_w(z, c)$ under treatment conditions (depth z and field size c) is the ratio of the doses measured in these conditions at the reference source-skin distance f'_R , with and without the wedge in the beam, for the same number of monitor units.

Using the same formalism as described in paragraph 6.2.2.2.1., the influence of the wedge filter on the dose can be expressed by the wedge factor in reference conditions $k'_w(c_R, z_R)$ weighted by the relative variation of this factor with field size and depth.

$$k'_w(z, c) = k'_w(z_R, c_R) \cdot a'_w(z_R, c) \cdot b'_w(z, c) \quad (6.14)$$

with

$$a'_w(z_R, c) = \frac{O'_R(c, w)}{O'_R(c)} \quad (6.15)$$

$$b'_w(z, c) = \frac{P_R(z, c, f'_R, w)}{P_R(z, c, f'_R)} \quad (6.16)$$

By substituting the expressions of a'_w and b'_w in equation 6.14, we obtain :

$$k'_w(z, c) = k'_w(z_R, c_R) \cdot \frac{O'_R(c, w)}{O'_R(c)} \cdot \frac{P_R(z, c, f'_R, w)}{P_R(z, c, f'_R)} \quad (6.17)$$

6.2.3.2.2. Dose calculation

In a manner similar to that used for the isocentric technique, the absorbed dose in the presence of the wedge can be calculated from the absorbed dose per monitor unit $\dot{D}'_R$ determined in reference conditions of the open beam :

- either by using the wedge factor in reference conditions $k'_w(c_R, z_R)$ with the depth dose data related to the wedged beam (figure 6-8).

$$D'(z, c, f'_R, w) = \dot{D}'_R \cdot U \cdot O'_R(c, w) \cdot k'_w(z_R, c_R) \cdot \frac{P_R(z, c, f'_R, w)}{100} \quad (6.18)$$

- or, by applying the wedge factor in treatment conditions $k'_w(z, c)$ and depth dose data of the corresponding open beam :

$$D'(z, c, f'_R, w) = \dot{D}'_R \cdot U \cdot O'_R(c) \cdot k'_w(z, c) \cdot \frac{P_R(z, c, f'_R)}{100} \quad (6.19)$$

The expression of $k'_w(z, c)$ is given by relation 6.14 or 6.17.

6.2.3.3. Relation between the wedge factor determined in a large water phantom and in a mini-phantom

Similarly to equation 6.10, the relation between the wedge factor determined in a large water phantom is related to the wedge factor in a mini-phantom according to :

$$k'_w(z_R, c_R) = k'_{o,w}(c_R) \cdot \frac{V(z_R, c'_R, w)}{V(z_R, c'_R)} \quad (6.20)$$

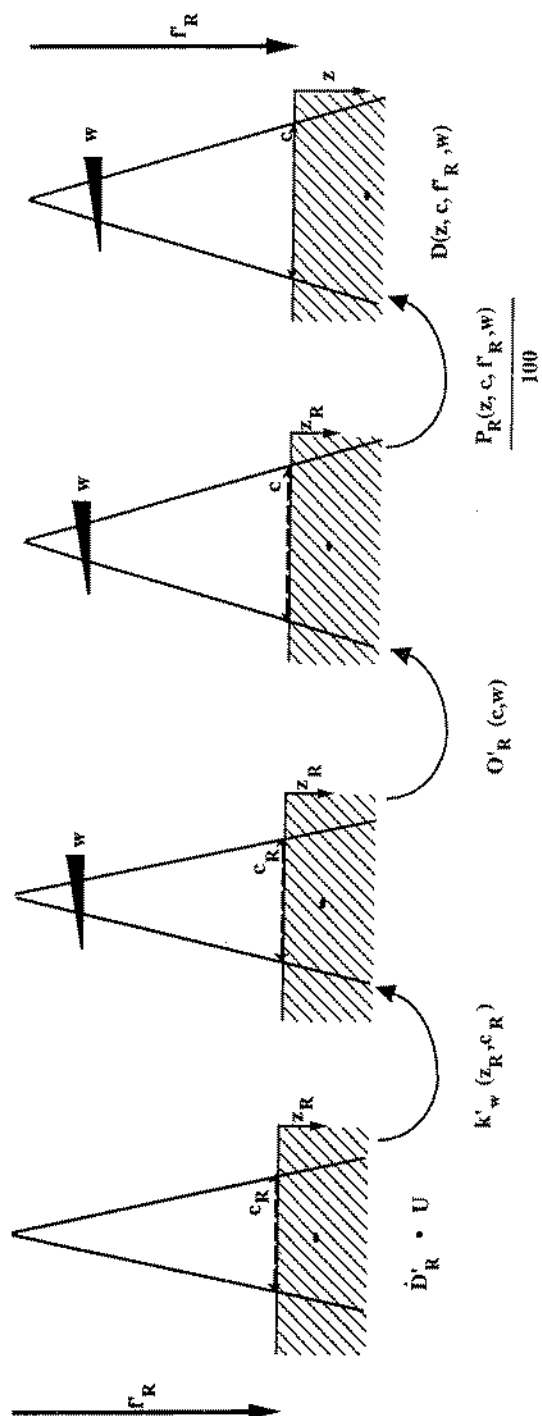


Figure 6-8. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a wedged beam from the dose per monitor unit under reference conditions in an open beam $\dot{D}'_R$ when the treatment is performed at the source-skin distance f'_R and when referring to measurements in a large water phantom.

6.3. BLOCKED BEAMS

6.3.1. General Concepts

In radiotherapy, it is generally necessary to treat patients with irregular fields in order to concentrate absorbed dose within the target volume and to minimise the dose to adjacent healthy tissue. In practice, this objective is achieved by interposing shielding blocks on a shadow tray at the exit side of the collimator or by using a multileaf collimator. Their use results in a modification of the incident photon fluence and consequently of the dose at the point of interest. The modification of the dose can be considered as a result of the following effects :

- partial loss of phantom scatter due to the shielding of part of the phantom volume by the blocks;
- attenuation of the photon fluence and production of electrons by the shadow tray used to support the blocks;
- modification in the number of photons and electrons scattered from the collimator, the flattening filter and the air between the source and the phantom surface;
- backscattered radiation from the blocks into the beam monitor chamber;
- scattered radiation produced in the blocks.

As a first approximation, the dose in a blocked field at a point located on the beam axis can be calculated from the dose measured under reference conditions in an open beam by taking into account the transmission of the shadow tray and the dose modifications linked to the reduction of field size by the shielding blocks, both being measured at the reference depth. The dose variation with depth, from the reference depth to the depth of interest, is considered by referring to dosimetric quantities (e.g., T , P_R .) applied to the actual irradiated field size defined by the blocks.

This simple approach allows the calculation of the dose in treatment conditions from the reference dose according to a formalism involving a set of factors and dosimetric parameters measured only in a large water phantom, i.e. both changes in the incident photon beam and modification of the scattering volume are simultaneously taken into account. In the case of blocked beams, both the primary and the scattered component of the dose are influenced in a complex way which cannot easily be predicted by simple relations.

A relatively large number of measurements is therefore needed in order to take into account the various configurations of blocked beams encountered in clinical practice. For this complex situation, it is simpler to consider separately the contribution of photons scattered from the treatment head and in the phantom. This can be achieved in practice by measurements performed both with a mini-phantom and a large phantom. The mini-phantom allows to determine the output ratio O_0 and the transmission factor of the shadow tray. The variation of O_0 depends only upon the collimator setting and the equivalent field size c_e should be calculated from the X and Y positions of the jaws (see section 5.6). If there is an influence of the blocks on O_0 , this influence should be determined in the mini-phantom.

The influence on the reference dose of photons scattered in the volume delimited by the blocks is quantified by the volume scatter ratio V . V is determined from measurements at the reference depth performed with the mini-phantom and in a large phantom. In practice, V can be measured for some typical square field sizes covering the scale of field sizes used in clinical practice (see section 8.5).

For the application to the complex field sizes delimited by the blocks, the equivalent field size s_b (equal to s_e) should be calculated using convenient computer software, by the application of Clarkson's method (see section 4.6.2) or by application of another published method. In some computer planning systems, the quantities depending on s_b (e.g. V , T , P) are calculated directly from basic beam data.

In the following sections, the two approaches for dose calculation in blocked beams are presented. The first detailed method is based on measurements performed with the mini-phantom and a large phantom. The second method is based on measurements performed in a large phantom only.

The proposed relationships for dose calculation for blocked fields apply to points located along the central axis. They are valid for any block configuration using blocks positioned outside this axis.

6.3.2. Isocentric Conditions

The reference conditions of depth z_R and distance between the source and the point of interest f_R are identical to those used for open beams (see chapter 5).

6.3.2.1. Dose calculation based on measurements with a mini-phantom.

6.3.2.1.1. Transmission factor of the shadow tray

The transmission factor $k_{o,t}(c)$ of the shadow tray is defined as the ratio of the doses measured with a mini-phantom under reference conditions (depth z_R and distance to the source f_R) and for the collimator opening c used for treatment, with and without the shadow tray, for the same number of monitor units.

$$k_{o,t}(c) = \frac{D_{o,t}(z_R, c, f_R)}{D_o(z_R, c, f_R)} \quad (6.21)$$

$k_{o,t}$ is assumed to be independent of the distance to the source provided that the two doses are measured at the same distance. $k_{o,t}$ depends slightly on the collimator opening because of the additional photons scattered by the shadow tray. This factor must be previously determined for the whole range of collimator openings used in clinical practice.

The transmission factor $k_{o,t}$ is defined at the reference depth z_R (figure 6-9). It is dependent on the depth of measurement. It will be closer to unity at shallower depths due to the contribution of electrons and photons scattered from the tray.

$k_{o,t}$ varies with the photon beam quality for any given shadow tray but should in principle be equal for a large phantom or a mini-phantom.

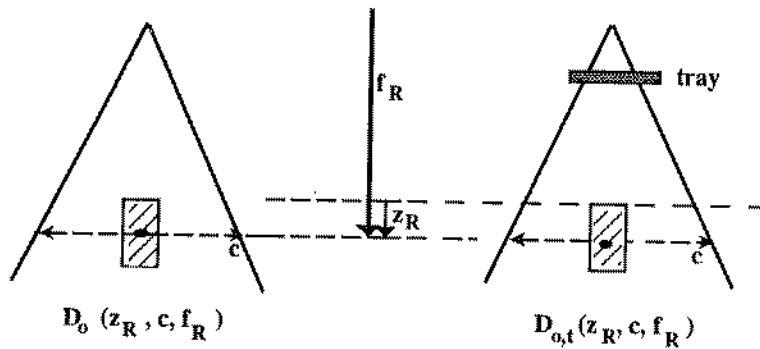


Figure 6-9. Diagram illustrating the definition of the transmission factor of the shadow tray $k_{o,t}$ measured in isocentric conditions with a mini-phantom.

6.3.2.1.2. Correction factor for modification of the dose by shielding blocks.

A factor $k_{o,b}(c, s_b)$ is defined as the ratio of the doses measured with a mini-phantom under reference conditions, depth z_R , source to point of interest distance f_R , for the collimator opening c , with and without the shielding blocks. The field size s_b corresponds to the field size defined by the blocks at f_R (figure 6-10).

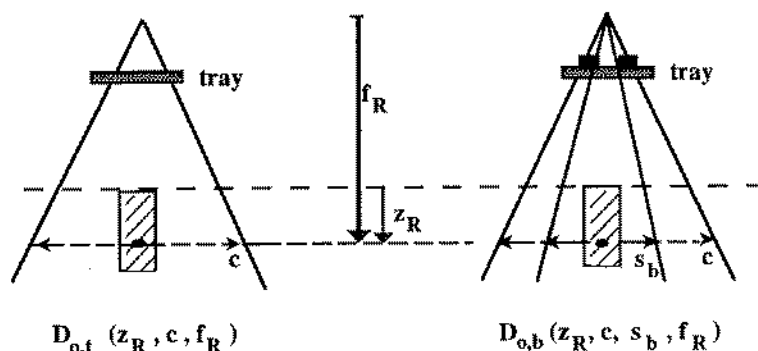


Figure 6-10. Diagram illustrating the definition of the correction factor $k_{o,b}$ taking into account the modifications in the dose due to the shielding blocks. The factor $k_{o,b}$ is assumed to be determined in isocentric conditions with a mini-phantom.

The doses must be measured with the shadow tray in place. $k_{o,b}$ includes a possible variation of $k_{o,t}$ with the irradiated area of the shadow tray.

$$k_{o,b}(c, s_b) = \frac{D_{o,b}(z_R, c, s_b, f_R)}{D_{o,t}(z_R, c, f_R)} \quad (6.22)$$

This ratio represents the variation of head scatter, when placing shielding blocks on the tray. For most photon beam energies, $k_{o,b}$ does not differ from 1 by more than 2%. However, $k_{o,b}$ has been found to present a variation up to 5 % for 25 MV X-rays when 80 % of the area of a 20 cm x 20 cm field size is shielded (Van Dam et al., 1992). Smaller variations of $k_{o,b}$ have been reported, for similar conditions, by other authors (Luxton and Astrahan, 1988, Tatcher and Bjärngard, 1994, Lam and Ten Haken, 1996).

6.3.2.1.3. Dose calculation.

The dose under treatment conditions $D(z, c, s_b, f_R)$ can be derived from the dose per monitor unit under reference conditions $\dot{D}_R$ according to the following relation (see figure 6-11).

$$D(z, c, s_b, f_R) = \dot{D}_R \cdot U \cdot O_p(c) \cdot \frac{V(z_R, s_b)}{V(z_R, c_R)} \cdot k_{o,t}(c) \cdot k_{o,b}(c, s_b) \cdot T_b(z, s_b) \quad (6.23)$$

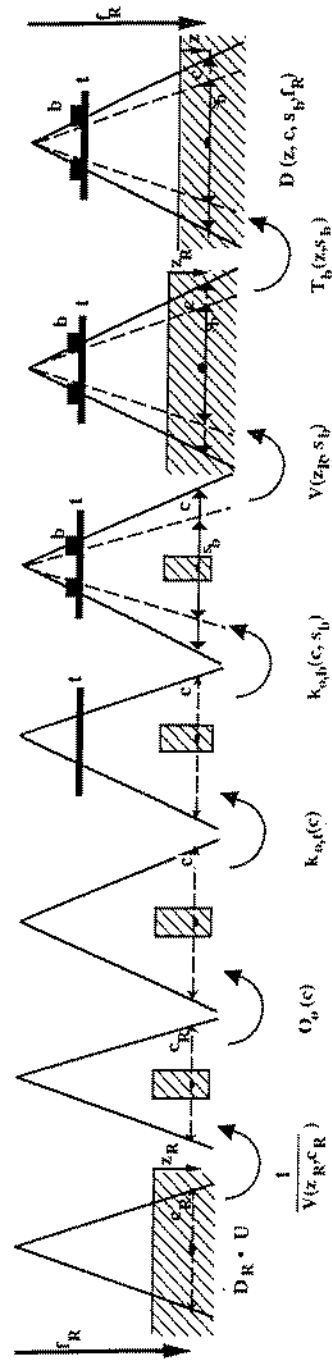


Figure 6-11. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a blocked beam from the dose per monitor unit under reference conditions D_R when the treatment is performed in isocentric conditions at the reference distance f_R and when referring to measurements with a mini-phantom.

where

- $O_0(c)$ represents the output ratio measured with the mini-phantom at the reference depth z_R for the field size c at the distance f_R from the source.
- $\frac{V(z_R, s_b)}{V(z_R, c_R)}$ represents the ratio of the volume scatter ratios for the field sizes s_b and c_R at f_R which is equal to $\frac{O_R(s_b)}{O_0(s_b)}$.
- $T_b(z, s_b)$ represents the tissue-phantom ratio measured with the shadow tray in the beam.

T values for z smaller than z_R can be modified by the presence of a tray and should be checked for the tray-to-skin distances in practical use.

The definitions of depths z and z_R , field sizes c_R , c and s_b , distance f_R are the same as those used in Chapter 5.

Relation 6.23 can be re-written as:

$$D(z, c, s_b, f_R) = \dot{D}_R \cdot U \cdot O_R(s_b) \cdot \frac{O_0(c)}{O_0(s_b)} \cdot k_{o,t}(c) \cdot k_{o,b}(c, s_b) \cdot T_b(z, s_b) \quad (6.24)$$

6.3.2.2. Dose calculation based on measurements in a large water phantom.

6.3.2.2.1. Transmission factor of the shadow tray.

The transmission factor k_t of the shadow tray is defined as the ratio of the doses measured in a large phantom at the reference depth z_R with and without the shadow tray, for the same number of monitor units (figure 6-12).

$$k_t(c) = \frac{D_t(z_R, c, f_R)}{D(z_R, c, f_R)} \quad (6.25)$$

As for $k_{o,t}$, k_t is assumed to be independent of the distance to the source provided that the two doses are measured at the same distance. k_t can depend slightly on the collimator opening. It also depends on the photon beam quality.

Typical values of k_t which have been obtained at the reference depth and in a 10 cm x 10 cm field size at the reference source-skin distance, for a shadow tray consisting of a plate of PMMA, 1 cm thick, are 0.93 in a Co-60 γ -ray beam, 0.945 for 5 MV x-rays ($T_{20/10} = 0.649$) and 0.97 for 18 MV x-rays ($T_{20/10} = 0.771$).

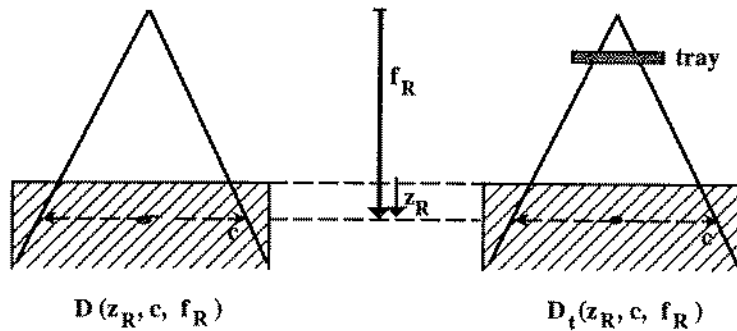


Figure 6-12. Diagram illustrating the definition of the transmission factor of the shadow tray k_t measured in isocentric conditions in a large water phantom.

6.3.2.2.2. Correction factor for modification of the dose by shielding blocks.

The factor k_b is defined as the ratio of the doses measured in a large phantom in reference conditions, depth z_R and distance to the source f_R , for the collimator opening c , with and without the shielding blocks, respectively (figure 6-13). The two doses must be measured with the shadow tray in place k_b includes a possible variation of k_t with the irradiated area of the shadow tray.

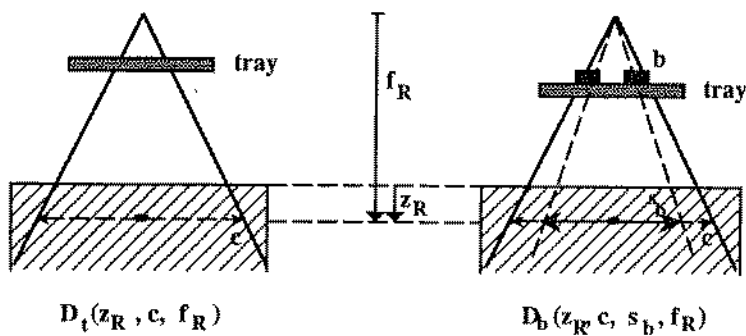


Figure 6-13. Diagram illustrating the definition of the correction factor k_b taking into account the modifications in the dose due to the presence of the shielding blocks. The factor k_b is assumed to be determined in isocentric conditions in a large water phantom.

$$k_b(c, s_b) = \frac{D_b(z_R, c, s_b, f_R)}{D_t(z_R, c, f_R)} \quad (6.26)$$

The correction factor $k_b(c, s_b)$ includes both changes in head and phantom scatter due to the presence of shielding blocks in the beam. Its determination, in practice, is not obvious because it necessitates a lot of measurements in order to take into account all configurations of blocked beams encountered in clinical practice.

Another way to obtain k_b values is to consider separately the effects of shielding blocks on head and phantom scatter and to refer to the relation between measurements in a large phantom and in a mini-phantom:

$$k_b(c, s_b) = k_{o,b}(c, s_b) \cdot \frac{k_{o,t}(c)}{k_t(c)} \cdot \frac{V(z_R, s_b)}{V(z_R, c)} \quad (6.27)$$

As a first approximation, the transmission factor of the shadow tray k_t can be considered to be independent of the type of phantom used for its determination (large phantom or mini-phantom). The variation of k_t values measured at depth z_R has been found to be less than 1 % for 10 cm x 10 cm and 20 cm x 20 cm field sizes both in ^{60}Co γ -ray beams and in 18 MV x-ray beams. For these situations k_t can be taken equal to $k_{o,t}$.

$\frac{V(z_R, s_b)}{V(z_R, c)}$ represents the ratio of volume scatter ratios at z_R for the field sizes s_b and c delimited by the blocks and the collimator at f_R , respectively.

The volume scatter ratios V are experimentally determined from measurements in a large phantom and a mini-phantom.

6.3.2.2.3. Dose Calculation

In isocentric conditions, the dose $D(z, c, s_b, f_R)$ at depth z for a collimator setting c , an irradiated square field size defined by the blocks s_b and distance to the source f_R , is related to the dose per monitor unit $\dot{D}_R$ in reference conditions (z_R, c_R, f_R) by the following equation :

$$D(z, c, s_b, f_R) = \dot{D}_R \cdot U \cdot O_R(c) \cdot k_t(c) \cdot k_b(c, s_b) \cdot T_b(z, s_b) \quad (6.28)$$

The different steps between $\dot{D}_R$ and D are shown in figure 6-14.

In this relation :

- $O_R(c)$ represents the output ratio measured in the large phantom at the reference depth z_R in the open beam for the field size c at the distance f_R from the source .

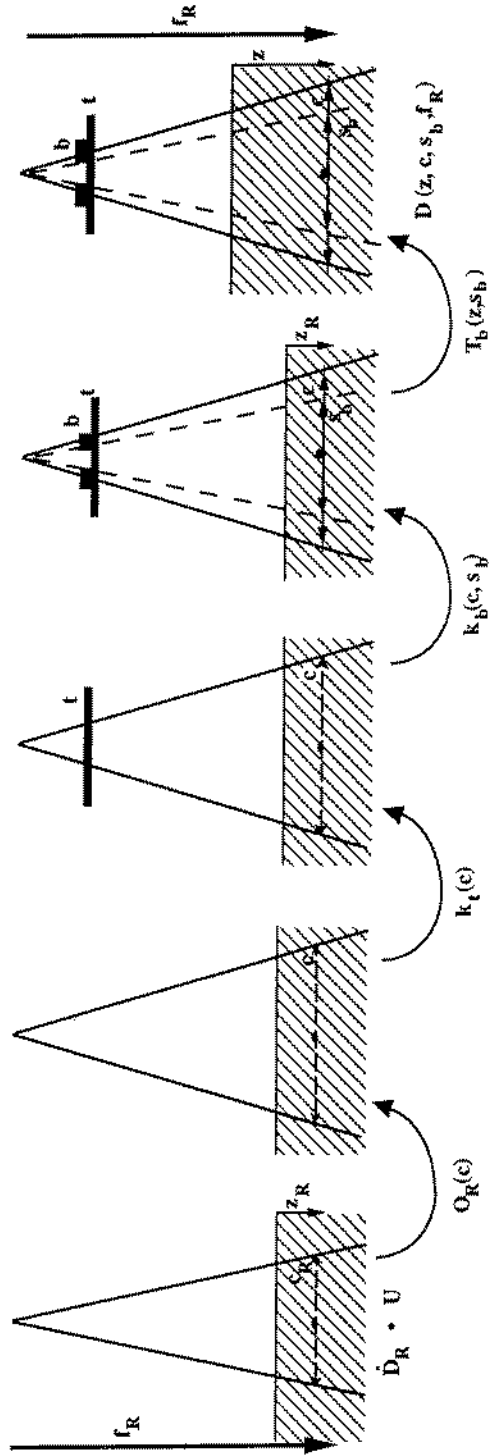


Figure 6-14. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a blocked beam from the dose per monitor unit $\dot{D}_R$ under reference conditions when the treatment is performed in isocentric conditions at the reference distance f_R and when referring to measurements in a large water phantom.

- $T_b(z, s_b)$ represents the tissue-phantom ratio measured with the shadow tray in the beam. It can be approximated by the tissue-phantom ratio measured in the open beam for the same field size when considering the depths z greater than the reference depth.

Substitution of equation 6.27 into equation 6.28 now yields:

$$D(z, c, s_b, f_R) = \dot{D}_R \cdot U \cdot O_R(c) \cdot \frac{V(z_R, s_b)}{V(z_R, c)} \cdot k_{o,t}(c) \cdot k_{o,b}(c, s_b) \cdot T_b(z, s_b) \quad (6.29)$$

A comparison of equations 6.24 and 6.29 shows that the equation 6.29 starts with the output ratio for the unblocked field and corrects for the difference in phantom scatter between the blocked and unblocked fields. Equation 6.24, on the other hand, starts with the output ratio for the blocked field and corrects then for the difference in head scatter for the field sizes corresponding to the blocked and unblocked fields.

6.3.3. Fixed Source-Skin Distance

When the treatments are performed at a fixed source-skin distance, the distance and the field are usually defined at the skin of the patient. In reference conditions, they are identified by the symbols f'_R and c_R , respectively (see section 5.5.1.1).

6.3.3.1. Dose calculation from measurements with a mini-phantom

6.3.3.1.1. The transmission factor of the shadow tray is given by :

$$k'_{o,t}(c) = \frac{D'_{o,t}(z_R, c, f'_R)}{D'_o(z_R, c, f'_R)} \quad (6.30)$$

$k'_{o,t}$, like $k_{o,t}$, is defined as the ratio of two doses measured at the same distance from the source (figure 6-15) and can be considered to be independent of the distance to the source, i.e. as a first approximation $k'_{o,t}(c) = k_{o,t}(c)$.

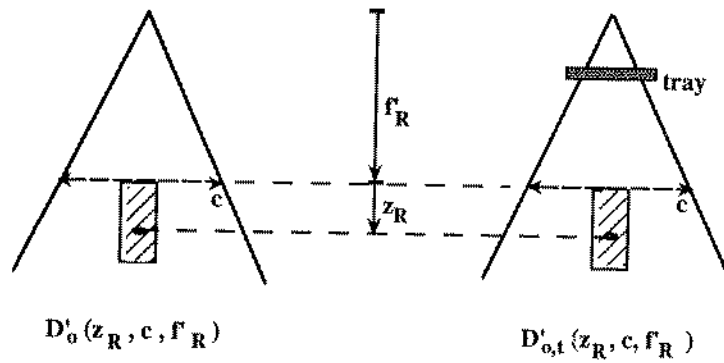


Figure 6-15. Diagram illustrating the definition of the transmission factor of the shadow tray $k'_{o,t}$ measured at the reference source-skin distance f'_R with a mini-phantom.

6.3.3.1.2. Correction factor for modification of the dose by the shielding blocks.

The factor $k'_{o,b}(c, s_b)$ is defined as the ratio of the doses measured with a mini-phantom under conditions (z_R, f'_R) for the collimator opening c , with and without the shielding blocks and shadow tray (figure 6-16). The field sizes c and s_b delimited by the blocks are defined at the distance between the source and the phantom surface f'_R .

$$k'_{o,b}(c, s_b) = \frac{D'_{o,b}(z_R, c, s_b, f'_R)}{D'_{o,t}(z_R, c, f'_R)} \quad (6.31)$$

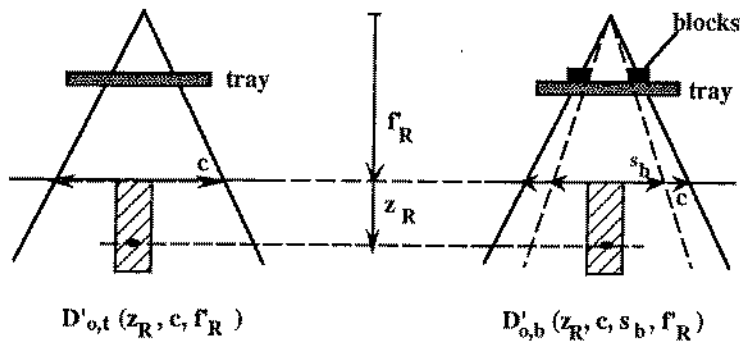


Figure 6-16. Diagram illustrating the definition of the correction factor $k'_{o,b}$ taking into account the modifications in the dose due to the presence of the shielding blocks. The factor $k'_{o,b}$ is assumed to be determined at the reference source-skin distance f'_R with a mini-phantom.

6.3.3.1.3. Dose calculation.

The relation between the dose under treatment conditions D' and the dose per monitor unit $\dot{D}'_R$ under reference conditions is (see figure 6-17):

$$D'(z, c, s_b, f'_R) = \dot{D}'_R \cdot U \cdot O'_0(c) \cdot \frac{V(z_R, s'_b)}{V(z_R, c'_R)} \cdot k'_{o,t}(c) \cdot k'_{o,b}(c, s_b) \cdot \frac{P_{R,b}(z, s_b, f'_R)}{100} \quad (6.32)$$

where :

- $O'_0(c)$ is the output ratio measured with the mini-phantom at z_R for the field size c at f'_R .
- c'_R and s'_b are equal to $c_R \cdot \frac{f'_R + z_R}{f'_R}$ and $s_b \cdot \frac{f'_R + z_R}{f'_R}$ respectively, c_R and s_b being defined at the reference distance f'_R .
- $\frac{V(z_R, s'_b)}{V(z_R, c'_R)}$ is equal to the ratio of output ratios : $\frac{O'_R(s_b)}{O'_R(s_b)}$

Relation 6.32 can be re-written as:

$$D'(z, c, s_b, f'_R) = \dot{D}'_R \cdot U \cdot O'_R(s_b) \cdot \frac{O'_0(c)}{O'_0(s_b)} \cdot k'_{o,t}(c) \cdot k'_{o,b}(c, s_b) \cdot \frac{P_{R,b}(z, s_b, f'_R)}{100} \quad (6.33)$$

6.3.3.2. Dose calculation based on measurements in a large water phantom

6.3.3.2.1. Transmission factor of the shadow tray

The transmission factor $k'_t(c)$ of the shadow tray is defined as the ratio of the dose values measured in a large phantom at the reference depth z_R with and without the shadow tray for the same number of monitor units (figure 6-18). The two dose values are assumed to be measured at a source-phantom surface distance f'_R and for the collimator setting c defined at this distance.

$$k'_t(c) = \frac{D'_t(z_R, c, f'_R)}{D'(z_R, c, f'_R)} \quad (6.34)$$

k'_t is assumed to be independent of the distance to the source in the usual range of treatment distances, i.e. $k'_t(c) = k_t(c)$.

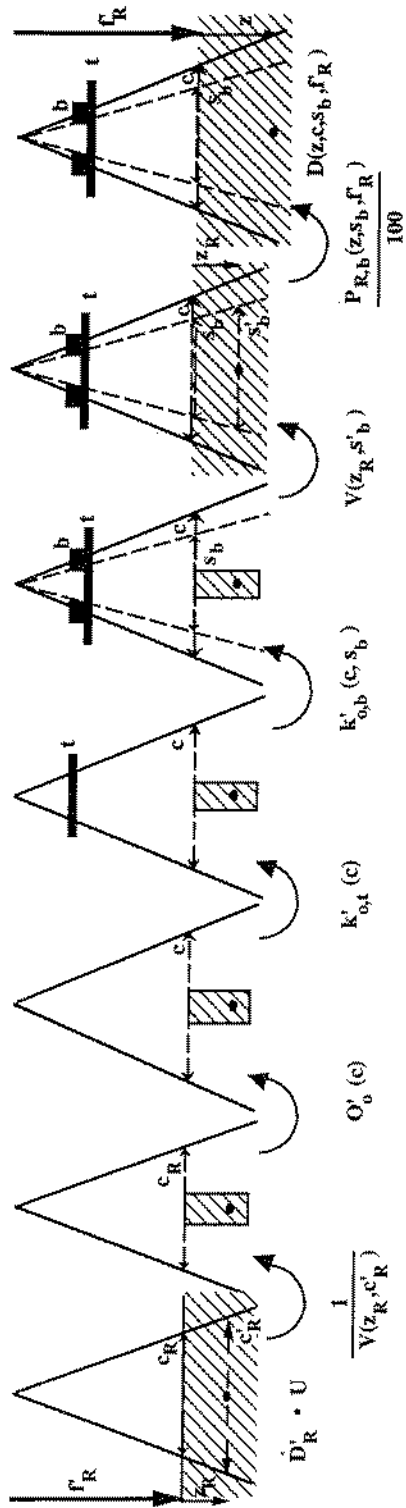


Figure 6-17. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a blocked beam from the dose per monitor unit $\dot{D}_R'$ under reference conditions when the treatment is performed at the reference source-skin distance f'_R and when referring to measurements with a mini-phantom.

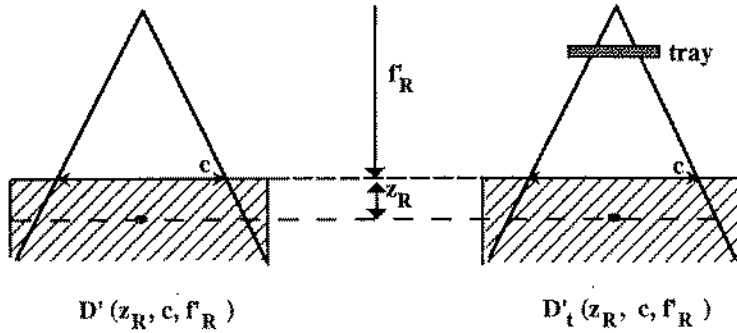


Figure 6-18. Diagram illustrating the definition of the transmission factor of the shadow tray k'_t measured at the reference source-skin distance f'_R in a large water phantom.

6.3.3.2.2. Correction factor for modification of the dose by shielding blocks.

The factor k'_b is defined as the ratio of the doses measured in the large phantom at the reference depth z_R , for the field c under consideration at f'_R , with and without shielding blocks (figure 6-19). The shadow tray must be used for these measurements.

$$k'_b(c, s_b) = \frac{D'_b(z_R, c, s_b, f'_R)}{D'_t(z_R, c, f'_R)} \quad (6.35)$$

The factor k'_b depends on the collimator opening and the position of the shielding blocks with respect to the beam axis.

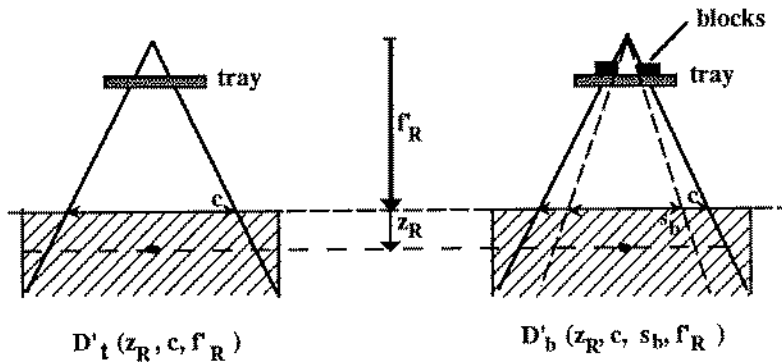


Figure 6-19. Diagram illustrating the definition of the correction factor k'_b taking into account the modifications in the dose due to the presence of the shielding blocks. The factor k'_b is assumed to be determined at the reference source-skin distance f'_R in a large water phantom.

As discussed for k_b (see section 6.3.2.2.3.), its determination in practice is not obvious and necessitates an important number of measurements in order to take into account the most common configurations of blocked fields used in clinical practice.

Nevertheless, a possibility to obtain k'_b values is to refer to some measurements performed with a mini-phantom and to apply the following equation :

$$k'_b(c, s_b) = k'_{o,b}(c, s_b) \cdot \frac{k'_{o,i}(c)}{k'_i(c)} \cdot \frac{V(z_R, s'_b)}{V(z_R, c')} \quad (6.36)$$

- as a first approximation, $k'_i(c) = k'_{o,i}(c)$

- $\frac{V(z_R, s'_b)}{V(z_R, c')}$ represents the ratio of volume scatter ratios at z_R , for the field sizes s'_b and c' delimited by the blocks and the collimator at $f'_R + z_R$, respectively.

6.3.3.2.3. Dose calculation

In fixed source-skin distance conditions, the dose $D'(z, c, s_b, f'_R)$ at a depth z , for a collimator setting c , an irradiated square field size defined by the blocks s and for a source-skin distance f'_R , can be calculated from the dose per monitor unit $\dot{D}'_R$ measured in reference conditions (z_R, c_R, f'_R) according to the equation :

$$D'(z, c, s_b, f'_R) = \dot{D}'_R \cdot U \cdot O'_R(c) \cdot k'_i(c) \cdot k'_b(c, s_b) \cdot \frac{P_{R,b}(z, s_b, f'_R)}{100} \quad (6.37)$$

All the dosimetric quantities and factors used in equation (6.37) refer to field sizes (c and s_b) defined at the source-skin distance f'_R .

The different steps used to derive $D'(z, c, s_b, f'_R)$ from $\dot{D}'_R$ are shown in figure 6-20.

Substitution of equation 6.36 into equation 6.37 yields:

$$D'(z, c, s_b, f'_R) = \dot{D}'_R \cdot U \cdot O'_R(c) \cdot \frac{V(z_R, s'_b)}{V(z_R, c')} \cdot k'_{o,i}(c) \cdot k'_{o,b}(c, s_b) \cdot \frac{P_{R,b}(z, s_b, f'_R)}{100} \quad (6.38)$$

6.4. COMBINED WEDGED AND BLOCKED BEAMS

When the treatment is performed with a wedged beam including shielding blocks, the calculation of the dose under treatment conditions from the reference dose per monitor unit is not straightforward. In this complex situation, it is recommended to refer to measurements performed with the mini-phantom in order to separate the

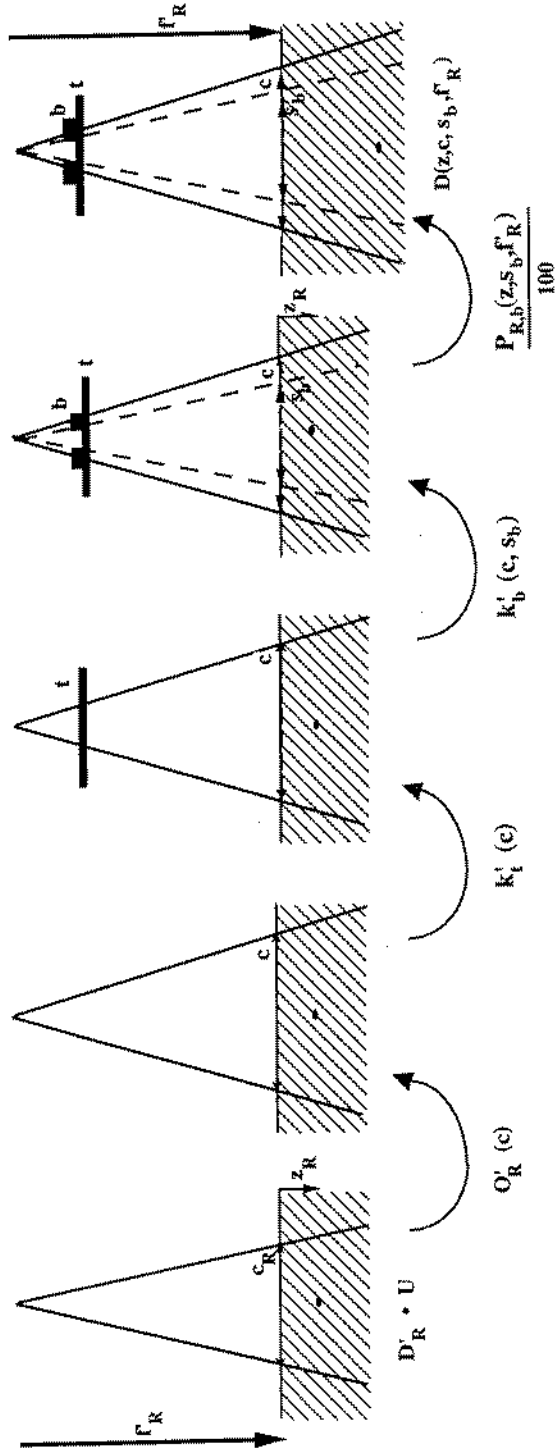


Figure 6-20. Diagram illustrating the different steps used for deriving the dose under treatment conditions in a blocked beam from the dose per monitor unit under reference conditions D'_R when the treatment is performed at the source-skin distance f'_R and when referring to measurements in a large water phantom.

contributions of head and phantom scatter linked to the different beam modifiers (wedge filter, shielding blocks).

According to paragraphs 6.2 and 6.3 and keeping the same definitions and symbols for the parameters as used before, the relations for the dose calculation are :

- Under isocentric conditions :

$$D(z, c, s_b, f_R, w) = \dot{D}_R \cdot U \cdot O_o(c, w) \cdot \frac{V(z_R, s_b, w)}{V(z_R, c_R)} \cdot k_{o,w}(c_R) \cdot k_{o,t}(c) \cdot k_{o,b}(c, s_b) \cdot T(z, s_b, w) \quad (6.39)$$

- Under fixed source-skin distance :

$$D'(z, c, s_b, f'_R, w) = \dot{D}'_R \cdot U \cdot O'_o(c, w) \cdot \frac{V(z_R, s'_b, w)}{V(z_R, c'_R)} \cdot k'_{o,w}(c_R) \cdot k'_{o,t}(c) \cdot k'_{o,b}(c, s_b) \cdot \frac{P_R(z, s_b, f'_R, w)}{100} \quad (6.40)$$

In the derivation of the equations, it is assumed that the modification of the reference conditions by introducing wedges or blocks in the open beam, can be handled independently.

7. MONITOR UNIT DETERMINATION FOR DISTANCES OTHER THAN THE REFERENCE DISTANCE

7.1. INTRODUCTION

Chapters 5 and 6 give the equations for dose calculations under treatment conditions for open and modified beams respectively, for a treatment distance equal to the reference distance. For treatments performed at a distance which is different from the reference distance, modifications arise in the primary photon fluence, (i.e. in the photon fluence coming directly from the target, scattered in the treatment head, and attenuated and scattered in the beam modifiers) and in the photons scattered in the irradiated volume. Two different approaches are considered to take distance variation into account.

The first approach is to adopt a separate set of reference conditions for this new treatment distance, which requires the determination of the various dosimetric parameters for this new distance. If this distance is not very different from the reference distance, then an approximation may be made by assuming that some parameters are independent of the distance. These parameters concern the ratios of two doses measured at the same distance from the source. These include output ratios O , volume scatter ratios V , tissue-phantom ratios T , transmission factors for the shadow tray k_t and shielding block factors k_b . Other factors must be remeasured. Using this approach, the standard formulae developed in chapters 5 and 6 may be used directly.

The second approach involves calculation of the distance dependent parameters from the reference data. This necessitates a complete separation of head and phantom scatter contributions, requiring measurements both in a mini-phantom and a large phantom. This second approach which relies on one set of reference conditions is the recommended one.

The present chapter reports the relations to be used for dose calculation in each of these two approaches. The relations are given for open, wedged and blocked beams.

7.2. MONITOR UNIT DETERMINATION USING MEASUREMENTS AT THE TREATMENT DISTANCE.

7.2.1. Reference Conditions for the new Treatment Distance

The source-phantom distance taken as reference distance is f'_T , corresponding to the source-skin distance used for the treatment.

s_R and z_R are the reference field size defined at f'_T and the reference depth, respectively.

$\dot{D}'_T(z_R, s_R, f'_T)$ is the dose per monitor unit measured in a large phantom under these conditions and is the new reference dose.

7.2.2. Open Beams

The dose under treatment conditions $D'_T(z, s, f'_T)$ can be obtained from $\dot{D}'_T$ according to a relation similar to relation 5.13:

$$D'_T(z, s, f'_T) = \dot{D}'_T \cdot U \cdot O'_T(s) \cdot \frac{P_R(z, s, f'_T)}{100} \quad (7.1)$$

where

- s is the treatment field size at the skin-source distance f'_T .
- $O'_T(s)$ is the output ratio measured in a large phantom under the treatment conditions.
- $\frac{P_R(z, s, f'_T)}{100}$ is the reference percentage depth dose at depth z , for the field size s at the source-skin distance f'_T .

If f'_T is not too different from f'_R , $P_R(z, s, f'_T)$ can be calculated from $P_R(z, s, f'_R)$ according to the relations given in Appendix 1.

7.2.3. Wedged Beams

The dose in treatment conditions with a wedge filter w in the beam $D'_T(z, s, f'_T, w)$ is related to $\dot{D}'_T$ by an expression similar to relation 6.18.

$$D'_T(z, s, f'_T, w) = \dot{D}'_T \cdot U \cdot O'_T(s, w) \cdot k'_w(z_R, s_R) \cdot \frac{P_R(z, s, f'_T, w)}{100} \quad (7.2)$$

where:

- s is the treatment field size at the source-skin distance f'_T .
- $k'_w(z_R, s_R)$ is the wedge factor measured in a large phantom at depth z_R for the reference field size s_R defined at f'_T .
- $O'_T(s, w)$ and $\frac{P_R(z, s, f'_T, w)}{100}$ are the output ratio and the reference percentage depth dose measured in a large phantom at the source-skin distance f'_T with the wedge filter in the beam. They can be related to the corresponding quantities in the unwedged beam by the relation 6.17, using the wedge factor $k'_w(z, s)$ instead of $k'_w(z_R, s_R)$.

7.2.4. Blocked Beams

The dose under treatment conditions with shielding blocks on a tray in the beam $D'_T(z, s, s_b, f'_T)$ can be derived from $\dot{D}'_T$ according to a relation similar to equation 6.37.

$$D'_T(z, s, s_b, f'_T) = \dot{D}'_T \cdot U \cdot O'_T(s) \cdot k'_i(s) \cdot k'_b(s, s_b) \cdot \frac{P_{R,b}(z, s_b, f'_T)}{100} \quad (7.3)$$

Where :

- s, s_b are the treatment field sizes at the source-skin distance f'_T defined by the collimator and the shielding blocks, respectively.
- $O'_T(s)$ is the output ratio measured for the field size s at f'_T in a large phantom.
- $k'_i(s)$ is the transmission factor of the shadow tray measured at f'_T (see Chapter 6).
- $k'_b(s, s_b)$ is a correction factor taking into account modification of the dose caused by the shielding blocks (see Chapter 6). The methods proposed for the determination of k'_i and k'_b are given in Chapter 6.
- $\frac{P_{R,b}(z, s_b, f'_T)}{100}$ is the reference percentage depth dose at depth z , for the field size s_b at the source-skin distance f'_T . It can be calculated from the reference percentage depth dose measured at f'_R according to the relation given Appendix 1.

By taking into account relationship 6.36 between $k'_t \bullet k'_b$ and $k'_{o,t} \bullet k'_{o,b}$, relation 7.3 can also be written as :

$$D'_T(z, s, s_b, f'_T) = \dot{D}_T \bullet U \bullet O'_T(s) \bullet \frac{V(z_R, s'_b)}{V(z_R, s')} \bullet k'_{o,t}(s) \bullet k'_{o,b}(s, s_b) \bullet \frac{P_{R,b}(z, s_b, f'_T)}{100} \quad (7.4)$$

where s'_b and s' are the field sizes at the distance $f'_T + z$ from the source and defined by the shielding blocks and the collimator, respectively.

7.3. MONITOR UNIT DETERMINATION USING MEASUREMENTS AT THE REFERENCE DISTANCE.

7.3.1. Use of Isocentric Conditions as Reference

The reference conditions are those defined at the usual treatment distance, which is different from the distance under consideration f'_T . The reference distance f_R is the distance between the source and the point at reference depth z_R (isocentric conditions). In this case, the reference dose per monitor unit $\dot{D}_R$ is measured at depth z_R with field size c_R at this level.

7.3.1.1. Open Beams

The dose under treatment conditions $D'_T(z, s, f'_T)$ is calculated from the reference dose $\dot{D}_T$ according to the following relations based on measurements performed with the mini-phantom (figure 7-1).

$$D'_T(z, s, f'_T) = \dot{D}_R \bullet U \bullet O_0(c) \bullet \left[\frac{f_R}{f'_T + z} \right]^2 \bullet \frac{V(z_R, s')}{V(z_R, c_R)} \bullet T(z, s') \quad (7.5)$$

In this relation :

- c and s' are the treatment field sizes defined by the collimator at f_R and $f'_T + z$, respectively
- $O_0(c)$ is the output ratio for field size c measured with the mini-phantom
- $\frac{V(z_R, s')}{V(z_R, c_R)}$ is the quotient of two volume scatter ratios

Assuming that V is independent of the distance from the source, as indicated in section 4.5.1, the ratio of V values can be considered, as a first approximation, to be equal to $\frac{O_R(s')}{O_o(s')}$.

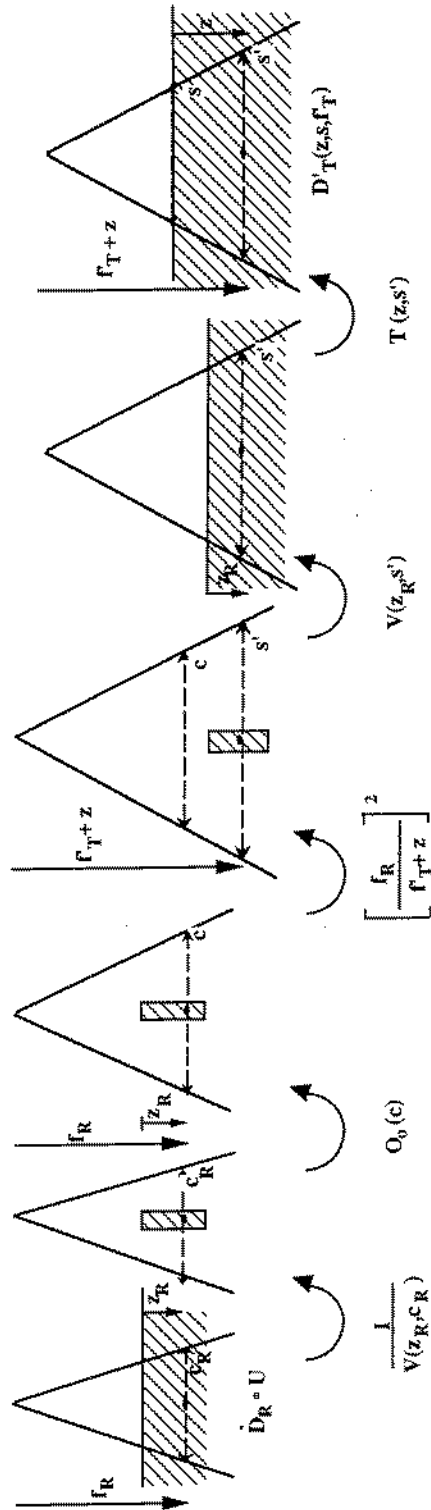


Figure 7-1. Diagram illustrating the different steps used for deriving the dose under treatment conditions in an open beam at a source-skin distance f'_T from the dose per monitor unit under reference conditions $\dot{D}_R$ measured at the reference distance f_R .

- $T(z, s')$ is the tissue-phantom ratio at depth z for field size s' . It is assumed to be independent of the distance from the source.

7.3.1.2. Wedged beams

According to figure 7-2, the relation between the dose under treatment conditions $\dot{D}_T$ and the reference dose per monitor unit $\dot{D}_R$ can be written :

$$\dot{D}_T(z, s, f'_T, w) = \dot{D}_R \cdot U \cdot O_0(c, w) \cdot \left[\frac{f_R}{f'_T + z} \right]^2 \cdot \frac{V(z_R, s', w)}{V(z_R, c_R)} \cdot k_{o,w}(c_R) \cdot T(z, s', w) \quad (7.6)$$

With :

- z : treatment depth
- z_R : reference depth
- c_R and c : reference and treatment field sizes defined by the collimator at f_R
- s and s' : treatment field sizes defined by the collimator at f'_T and $f'_T + z$

In relation 7.6,

- $O_0(c, w)$ represents the output ratio for field size c measured with a mini-phantom at f_R with the wedge filter in the beam.
- $k_{o,w}(c_R)$ is the wedge factor measured with a mini-phantom for the field size c_R at f_R . The factor $k_{o,w}(c_R)$ is related to the wedge factor $k_w(z_R, c_R)$ measured in reference conditions (z_R, c_R, f_R) in a large phantom according to the relation :

$$k_{o,w}(c_R) = k_w(z_R, c_R) \cdot \frac{V(z_R, c_R)}{V(z_R, c_R, w)} \quad (7.7)$$

where :

$V(z_R, c_R)$ and $V(z_R, c_R, w)$ represent the volume scatter factors measured in reference conditions without and with the wedge filter in the beam, respectively.

By combining relation 7.7 and the following relation between the output ratios O_R , O_0 and V :

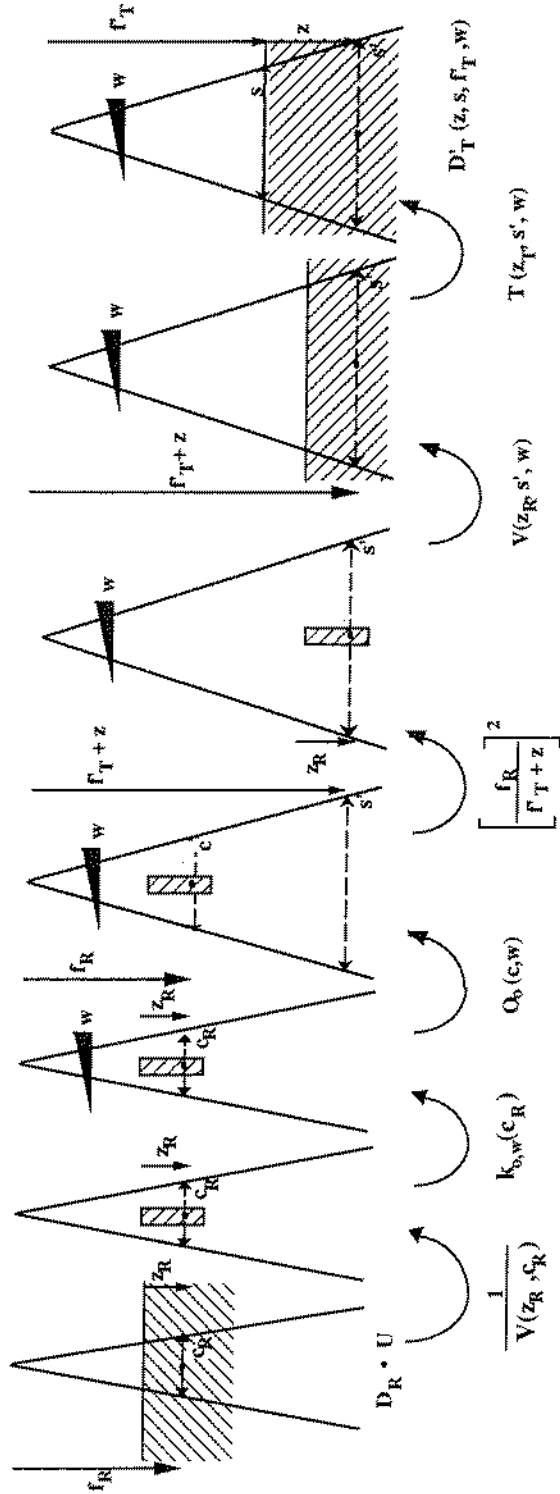


Figure 7-2. Diagram illustrating the different steps for deriving the dose under treatment conditions in a wedged beam at a source-skin distance f_T' from the dose per monitor unit under reference conditions $\dot{D}_R$ measured at the reference distance f_R

$$\frac{V(z_R, s', w)}{V(z_R, c_R, w)} = \frac{O_R(s', w)}{O_o(s', w)} \quad (7.8)$$

relation 7.6 can be also written as :

$$D'_T(z, s, f_T, w) = \dot{D}_R \cdot U \cdot O_R(s', w) \cdot \frac{O_o(c, w)}{O_o(s', w)} \cdot \left[\frac{f_R}{f'_T + z} \right]^2 \cdot k_w(z_R, c_R) \cdot T(z, s', w) \quad (7.9)$$

The output ratios O_R , O_o , the wedge factor k_w and the tissue-phantom ratio T are assumed to be measured in the wedged beam at the reference distance between the source and the point of interest equal to f_R .

7.3.1.3. Blocked beams

According to figure 7-3, the relation between the dose under treatment conditions D'_T and the reference dose per monitor unit $\dot{D}_R$ can be written as :

$$D'_T(z, s, s_b, f'_T) = \dot{D}_R \cdot U \cdot O_o(c) \cdot \left[\frac{f_R}{f'_T + z} \right]^2 \cdot \frac{V(z_R, s'_b)}{V(z_R, c_R)} \cdot k_{o,t}(s') \cdot k_{o,b}(s', s'_b) \cdot T_b(z, s'_b) \quad (7.10)$$

with

- z : treatment depth
- z_R : reference depth
- c_R : reference field size defined by the collimator at f_R
- s and s' : field sizes defined by the collimator at f'_T and $f'_T + z$
- s_b and s'_b : field sizes defined by the shielding blocks at f'_T and $f'_T + z$.

In relation 7.10 :

- O_o , $k_{o,t}$, and $k_{o,b}$ refer to the output ratio, the transmission factor of the shadow tray and the correction factor for dose modification induced by shielding blocks, measured with a mini-phantom.
- $\frac{V(z_R, s'_b)}{V(z_R, c_R)}$ represents the ratio of two volume scatter ratios. As a first approximation, it can be considered equal to $\frac{O_R(s'_b)}{O_o(s'_b)}$

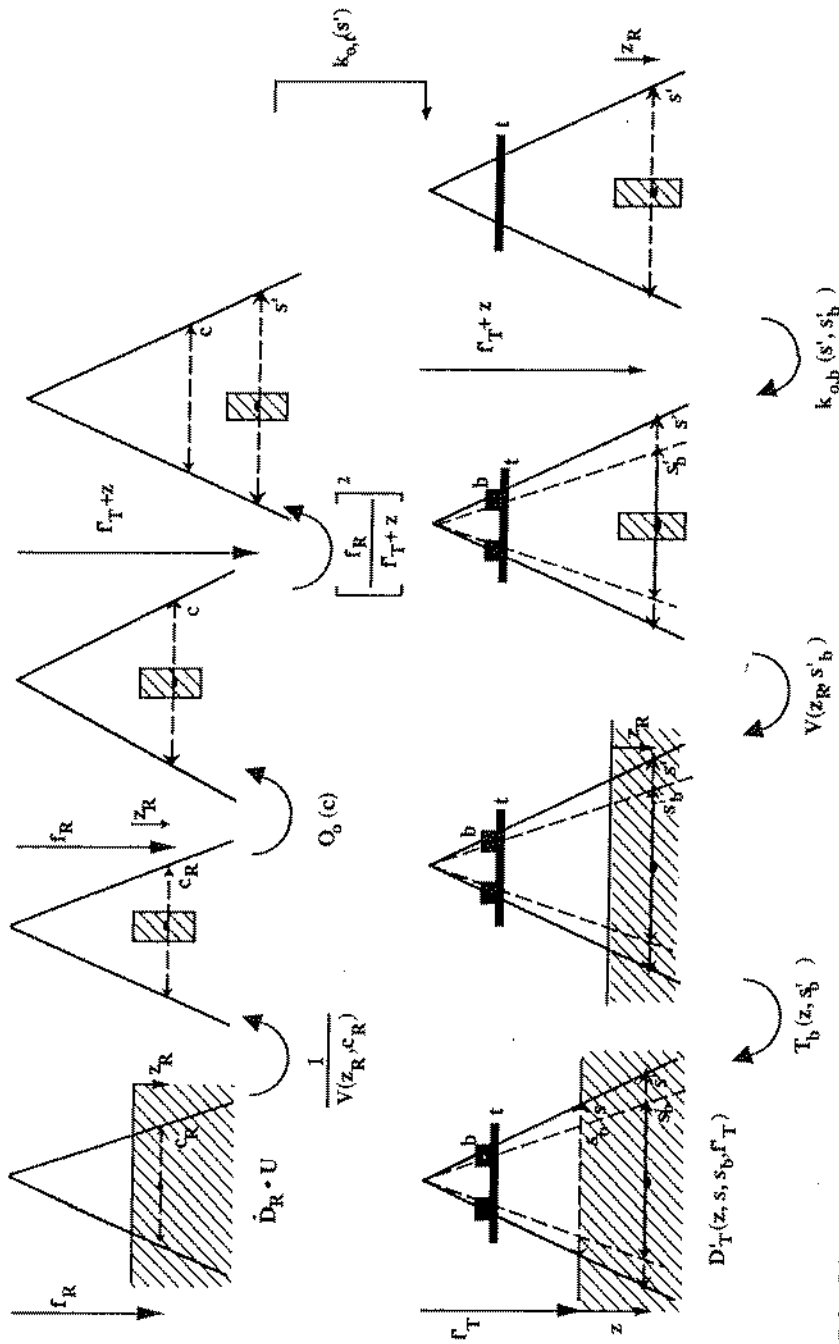


Figure 7-3. Diagram illustrating the different steps deriving the dose under treatment conditions in a blocked beam at a source-skin distance f'_T from the dose per monitor unit under reference conditions $\dot{D}_R$ measured at the reference distance f_R .

Based on these considerations, relation 7.10 can be rewritten as :

$$D'_T(z, s, s_b, f'_T) = \dot{D}_R \cdot U \cdot O_R(s'_b) \cdot \left[\frac{f_R}{f'_T + z} \right]^2 \cdot \frac{O_0(c)}{O_0(s'_b)} \cdot k_{o,i}(s') \cdot k_{o,b}(s', s'_b) \cdot T_b(z, s'_b) \quad (7.11)$$

The tissue-phantom ratio T_b , the output ratios O_0 and O_R are considered to be measured at the reference depth z_R and at the reference distance f_R . They are assumed to be independent of the distance from the source.

7.3.2. Use of Fixed Source-Skin Distance Measurements as Reference

The reference distance f'_R is the distance between the source and the phantom surface, the reference field size being defined at this level. In this situation, the reference dose per monitor unit, measured at depth z_R , is $\dot{D}'_R$.

The treatment distance is f'_T which is different from f'_R .

7.3.2.1. Open beams

The relation between the dose under treatment conditions $D'_T(z, s, f'_T)$ and the reference dose $\dot{D}'_R$ is (figure 7-4):

$$D'_T(z, s, f'_T) = \dot{D}_R \cdot U \cdot O'_0(c) \cdot \left[\frac{f'_R + z_R}{f'_T + z_R} \right]^2 \cdot \frac{V(z_R, s'')}{V(z_R, c'_R)} \cdot \frac{P_R(z, s, f'_T)}{100} \quad (7.12)$$

where :

- c'_R is the reference field size at $f'_R + z_R$
- s'' is the treatment field size defined by the collimator at $f'_T + z_R$
- $O'_0(c)$ is the output factor for the field size c at f'_R measured with the mini-phantom
- $\frac{V(z_R, s'')}{V(z_R, c'_R)}$ is the quotient of two volume scatter ratios. As a first approximation, it can be taken equal to $\frac{O'_R(s)}{O'_0(s)}$

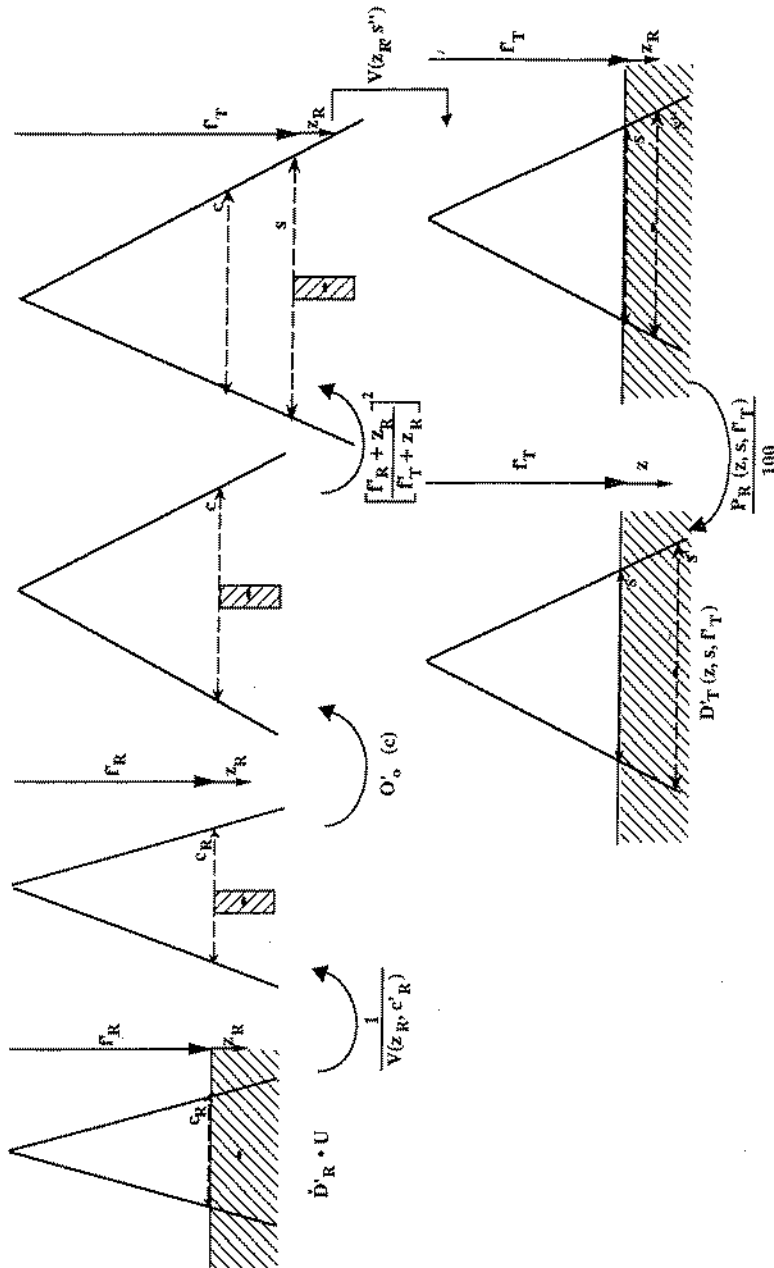


Figure 7-4. Diagram illustrating the different steps for deriving the dose under treatment conditions in an open beam at a source-skin distance f_T' from the dose per monitor unit under reference conditions $\dot{D}_R'$ measured at the reference source-skin distance f_R' .

- $\frac{P_R(z, s, f'_T)}{100}$ is the reference percentage depth dose at depth z , for the field size s at the source-skin distance f'_T . It can be calculated from $P_R(z, s, f'_R)$ according to the relation given in Appendix A.1

7.3.2.2. Wedged beams

The relation between the dose under treatment conditions $D'_T(z, s, f'_T, w)$ with the wedge filter in the beam and the dose under reference conditions $\dot{D}'_R(z_R, s_R, f'_R)$ is (figure 7-5).

$$D'_T(z, s, f'_T, w) = \dot{D}'_R \cdot U \cdot O'_0(c, w) \cdot \left[\frac{f'_R + z_R}{f'_T + z_R} \right]^2 \cdot \frac{V(z_R, s'', w)}{V(z_R, c'_R)} \cdot k'_{o, w}(c_R) \cdot \frac{P_R(z, s, f'_T, w)}{100} \quad (7.13)$$

where :

- z_R and z are the reference and treatment depths,
- c_R and c'_R are the reference field sizes defined by the collimators at f'_R and $f'_R + z_R$,
- s and s'' are the treatment field sizes defined by the collimator at f'_T and $f'_T + z_R$,
- c is the treatment field size corresponding to s , defined by the collimator at f'_R ,
- $k'_{o, w}(c_R)$ is the wedge factor measured with a mini-phantom for the field size c_R at f'_R ,
- $O'_0(c, w)$ is the output ratio measured with a mini-phantom for the field size c at f'_R with the wedge filter in the beam.

According to the same considerations as presented in paragraph 7.3.1.2., we can write :

$$k'_{o, w}(c_R) = k'_w(z_R, c_R) \cdot \frac{V(z_R, c'_R)}{V(z_R, c'_R, w)} \quad (7.14)$$

$$\text{and} \quad \frac{V(z_R, s'', w)}{V(z_R, c'_R, w)} = \frac{O'_R(s, w)}{O'_0(s, w)} \quad (7.15)$$

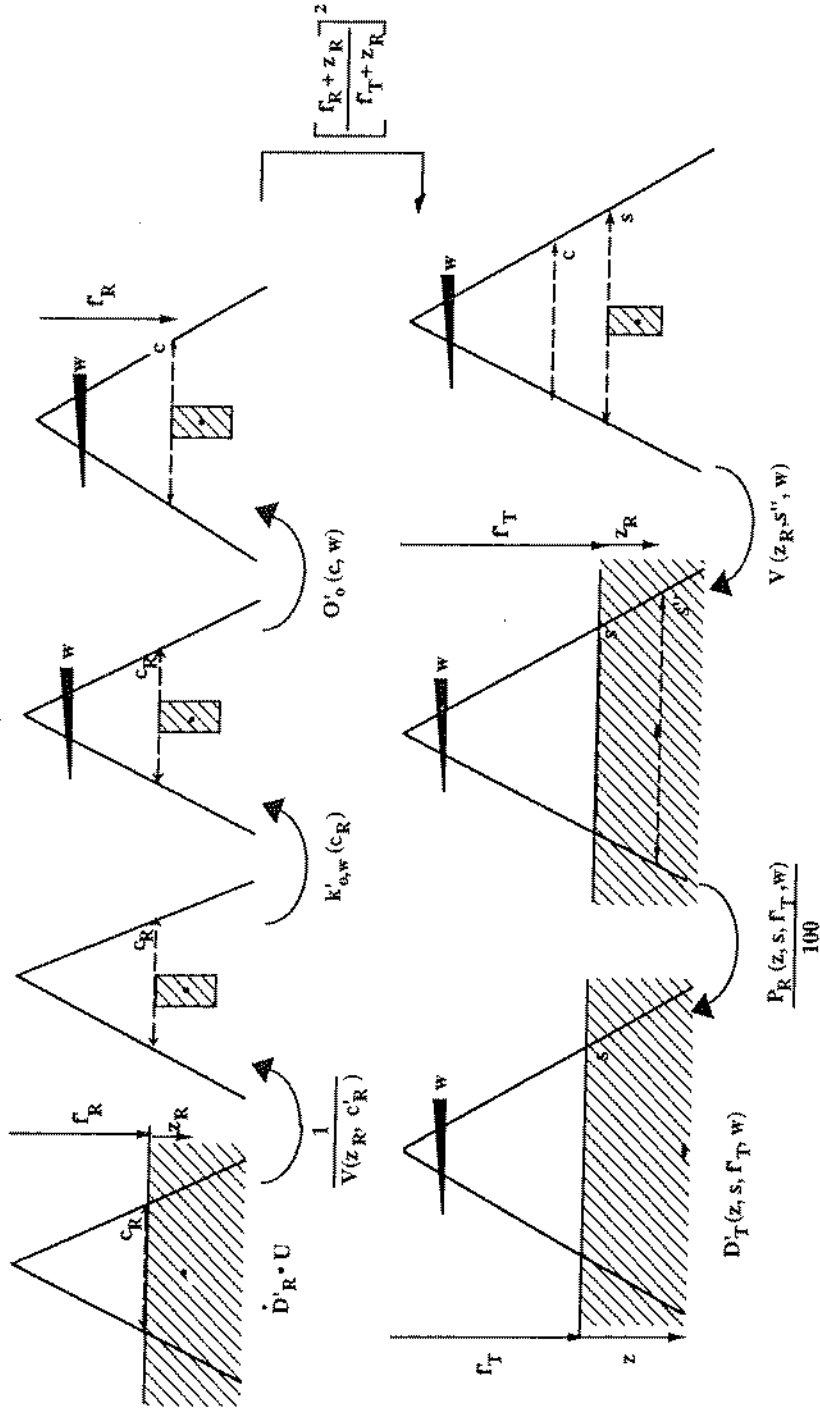


Figure 7-5. Diagram illustrating the different steps for deriving the dose under treatment conditions in a wedged beam at a source-skin distance f'_T from the dose per monitor unit under reference conditions D'_R measured at the reference source-skin distance f'_R .

By substituting equations 7.14 and 7.15 in equation 7.13, one obtains the general relation between the dose D'_T in treatment conditions and the dose in reference conditions $\dot{D}'_R$:

$$D'_T(z, s, f'_T, w) = \dot{D}'_R \cdot U \cdot O'_R(s, w) \cdot \frac{O'_0(c, w)}{O'_0(s, w)} \cdot \left[\frac{f'_R + z_R}{f'_T + z_R} \right]^2 \cdot k'_w(z_R, c_R) \cdot \frac{P_R(z, s, f'_T, w)}{100} \quad (7.16)$$

where :

- O'_R and O'_0 represent the output ratio in a large phantom and in a mini-phantom at depth z_R for the field size under consideration and with the wedge filter in the beam.
- $k'_w(z_R, c_R)$ is the wedge factor measured in a large phantom at z_R and for the reference field size c_R at f'_R .

The output ratios O'_0 , O'_R and the wedge factor k'_w are assumed to be measured in the wedged beam at the reference source-phantom surface f'_R .

The percentage depth dose P is assumed to be measured at the treatment distance under consideration f'_T or to be calculated from the percentage depth dose measured in the wedged beam at the reference distance f'_R according to the relations given in the Appendix A-1.

7.3.2.3. Blocked beams

The relation between the dose in treatment conditions $D'_T(z, s, s_b, f'_T)$ and the reference dose is $\dot{D}'_R$ (figure 7-6) :

$$D'_T(z, s, s_b, f'_T) = \dot{D}'_R \cdot U \cdot O'_0(c, w) \cdot \left[\frac{f'_R + z_R}{f'_T + z_R} \right]^2 \cdot \frac{V(z_R, s''_b)}{V(z_R, c'_R)} \cdot k'_{o,t}(s) \cdot k'_{o,b}(s, s_b) \cdot \frac{P_{R,b}(z, s_b, f'_T)}{100} \quad (7.17)$$

with :

- z : treatment depth
- z_R : reference depth
- c_R and c'_R : reference field sizes defined by the collimator at f'_R and $f'_R + z_R$.

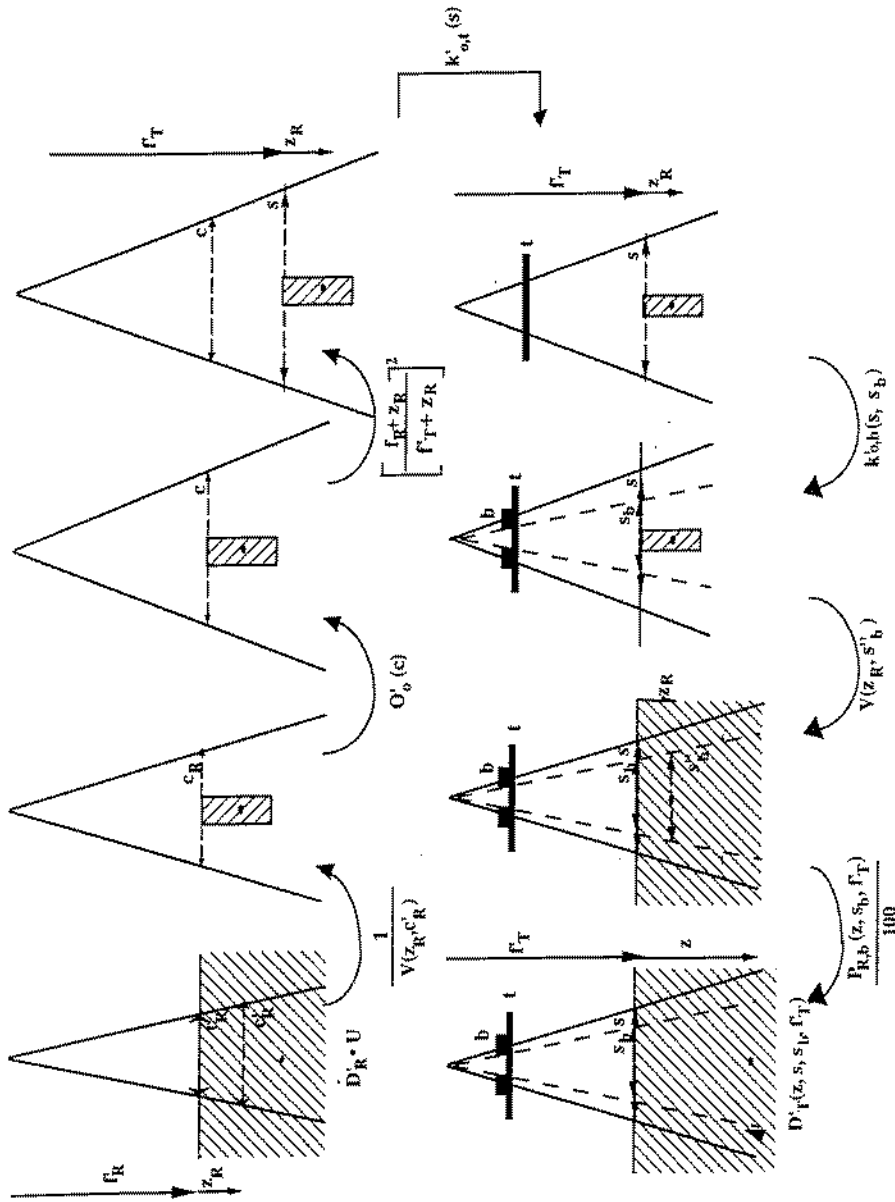


Figure 7-6. Diagram illustrating the different steps for deriving the dose under treatment conditions in a blocked beam at a source-skin distance f'_T from the dose per monitor unit under reference conditions $\dot{D}_R$ measured at the reference source-skin distance f'_R .

- c : field size defined by the collimator at f'_R and corresponding to the collimator setting during treatment
- s : field size defined by the collimator at the treatment source-skin distance f'_T
- s_b and s''_b : treatment field sizes defined by the shielding blocks at f'_T and $f'_T + z_R$, respectively
- O'_0 , $k'_{o,t}$ and $k'_{o,b}$: the output ratio, the transmission factor of the shadow tray and the correction factor taking into account the presence of shielding blocks in the beam; all measured with a mini-phantom.

By replacing the quotient of the volume scatter ratios by $\frac{O'_R}{O'_0}$, the relation between D'_T and D'_R can be written as:

$$D'_T(z, s, s_b, f'_T) = D'_R \cdot U \cdot O'_R(s_b) \cdot \frac{O'_0(c)}{O'_0(s_b)} \cdot \left[\frac{f'_R + z_R}{f'_T + z_R} \right]^2 \cdot k'_{o,t}(s) \cdot k'_{o,b}(s, s_b) \cdot \frac{P_{R,b}(z, s_b, f_R)}{100} \quad (7.18)$$

The output ratio O , the correction factors k_t and k_b and the volume scatter factors V are assumed to be independent of the distance from the source. They can be measured at the reference distance f'_R .

In this method, the percentage depth dose $P_{R,b}$ is assumed to be measured at the treatment distance under consideration f_R or to be calculated from the percentage depth dose P_R measured at the reference distance f'_R according to the relations given in Appendix A.1.

8. MEASUREMENTS

8.1. INTRODUCTION

The main purpose of this chapter is to describe how to measure the various physical quantities discussed in this report. These quantities concern the output per monitor unit of the treatment machine under reference conditions and a number of relative quantities required for the various MU calculation procedures. Each of these measurements requires special instrumentation, well described measurement procedures, specific phantom materials and quality control procedures. Although ample information on dose determinations under reference conditions can be found in the various dosimetry protocols and quality assurance documents that are already available, it is useful to summarize in this chapter in more detail those procedures and data required for the experimental determination of the physical quantities required for MU computations.

8.2. DOSIMETERS

8.2.1. Dosimeters for Dose Determinations under Reference Conditions

An ionization chamber in combination with an electrometer is the only dosimetry system recommended in modern dosimetry protocols to be used for dose determinations under reference conditions. Each radiotherapy centre generally has access to a number of dosimeters used for such a "beam calibration". At least one of these should be designated as the local reference dosimeter used solely to calibrate the other instruments. The number and type of chamber allowed as local standard in a particular dosimetry protocol varies therefore considerably as discussed elsewhere (Mijnheer et al., 1993). Each local reference chamber will have a calibration factor traceable to a primary or secondary standard and will be recalibrated at regular intervals (for example, every 2 years). The certificate of calibration will specify various performance elements, such as leakage rate, zero setting accuracy and dose linearity, as well as:

- energy specification of each calibration beam,
- the resultant calibration factor (exposure, air kerma or absorbed dose to water per instrument reading),
- the atmospheric conditions to which the calibration applies.

The local reference chamber should be kept in a secure, well-controlled environment. It should only be used by experienced staff. Field instruments are those used routinely within the centre for the measurement and calibration of radiation beams. Each field instrument used for calibration must itself be calibrated against the local standard, and should be recalibrated at regular intervals (e.g. every year).

The conversion of a measurement in a hospital using a field instrument at a specific photon beam quality, to absorbed dose to water using a calibration factor obtained from a standards dosimetry laboratory is a non-trivial problem. This procedure is complicated because most standards laboratories do not have the same beam qualities as used in radiotherapy centres. Therefore generally the starting point of a beam calibration is a ^{60}Co air kerma or exposure calibration factor. In some standards dosimetry laboratories absorbed dose to water standards for ^{60}Co gamma-rays and high-energy X-ray beams are becoming available. Such a procedure facilitates the beam calibration in a hospital, but these facilities are not yet generally available. IAEA TRS 277 and several national dosimetry protocols describe in detail how the transfer of calibration factors to the local instruments and the conversion of ionization chamber reading to absorbed dose to water should be performed for different types of ionization chamber. By following the procedures recommended in these codes of practice or protocols in detail, the calibration factor of the field instrument is traceable to an appropriate primary standard.

8.2.2. Dosimeters for Relative Dose Determinations

Most quantities discussed in this report require only relative dose measurements. For instance output ratios, wedge factors and tray factors are generally determined as the ratio of two readings of an ionization chamber. Most types of cylindrical air ionization chambers applied for absolute dose measurements are also suitable for this type of measurement. Also percentage depth dose curves or tissue-phantom ratio data can in principle be measured using such a type of field instrument. For relative measurements performed with cylindrical ionization chambers the effective point of measurement of the chamber should be taken at a fraction of the chamber radius in front of its centre (e.g. see IAEA, 1987).

In order to measure the large number of relative dose values required in clinical practice, water phantoms are available in which an ionization chamber can be scanned in two or three dimensions. Many manufacturers that produce these devices provide, in addition, small ionization chambers especially suited for relative dose measurements. These thimble type chambers have an inner diameter of about 4 mm

and a length between 10 and 15 mm. The chambers are either waterproof or are encapsulated in waterproof rubber. Because the dimensions and composition of the wall material is generally not known exactly, these chambers are only suitable for relative measurements.

Plane-parallel chambers are mainly used for the absorbed dose determination in low-energy electron and X-ray beams. They might, in principle, also be used for relative dose measurements in high-energy photon beams but are unsuitable for "beam calibration" because reliable data on wall correction factors (k_{att} , k_m , p_{wall}) are not available for most types of parallel-plate ionization chamber. Some of these chambers have very thin entrance walls which make them particularly useful for measurements in the build-up region. A review on the use of plane-parallel chambers has recently been given by Nahum and Thwaites (1993).

It should be noted that ratios of ionization chamber readings are only equal to ratios of dose values if the stopping power ratio water-to-air and perturbation correction factors are independent of field size and depth in the phantom. It has been shown that this is a valid assumption in clinical dosimetry except close to the surface in the build-up region.

Diodes are also applied for relative dose measurements. Due to their energy dependence it is difficult to predict if changes in diode readings are equal to dose variations. For that reason the back of some detectors has been covered with tungsten/epoxy layer to shield the diode from low-energy back-scattered photons. Such a shielding is particularly important in low-energy beams and at greater depths in phantoms. Percentage depth doses obtained with such an energy-compensated diode do agree generally within reasonable limits with ionization chamber data over a wide range of radiation qualities, field sizes and depths. Also for off-axis ratio measurements, both in air and in phantom, these diodes give reasonable agreement with ionization chamber data. The use of diodes in a beam of an accelerator applying a high frequency field might cause unexpected readings due to disturbances by the RF field. The magnitude of these deviating readings depends on the time constant of the electrometer, i.e. if the integration is too fast. For accurate dose measurements, a comparison of diode readings with ionization chamber measurements is always required for some extreme geometries and might reveal unexpected deviations (Karlsson et al., 1997).

Film dosimetry is an attractive alternative to ionization chambers and diodes. Film has a high spatial resolution, it can be inserted into radiation fields with a minimum perturbation and only short accelerator beam times are required. However,

film response varies with many parameters such as film processing procedure and variation in photon beam quality with depth, field size and irradiation geometry. For that reason film has to be calibrated under circumstances similar to those under the actual irradiation conditions. If, however, the necessary precautions and corrections are applied, film dosimetry is a simple, accurate and fast method to measure relative dose distributions for a range of megavoltage photon beam energies.

A review of the various methods for relative dose determination can be found at several places (e.g. Khan, 1984, Horton, 1993).

8.2.3. Operational Corrections of Ionization Chamber Readings

In order to derive the absorbed dose to water in a phantom from an ionization chamber reading, a number of correction factors has to be applied to obtain the same reference conditions during calibration and phantom measurement. These operational corrections concern the effect of temperature, pressure, relative humidity, ion recombination and polarity on the ionization chamber reading.

Temperature and pressure.

As atmospheric conditions change, a temperature and pressure correction must be applied to the reading in order to account for the difference relative to the values stated in the calibration certificate. It is therefore essential in performing measurements to ensure that equipment has had sufficient time to reach equilibrium conditions. In particular the relevant temperature is that of the phantom in which the chamber is to be used and the relevant pressure is the actual pressure at the place of measurement.

There are some situations where corrections for temperature and pressure are not appropriate, e.g. when using ionization chambers which are intrinsically sealed, with no gas exchange with the surrounding atmosphere. Some instruments correct automatically for temperature and pressure changes for instance by using internal temperature and pressure transducers. Care must be taken with instruments of these types to ensure that they are either totally sealed or that any correction has been applied correctly.

Humidity

The response of an ionization chamber is affected by the amount of water vapour within the collecting volume which affects both the mass of air present and the W/e value. The magnitude of variation is small, less than 0.3%, in comparison to

the other environmental effects. Hence, for a chamber calibrated by a standards laboratory at a relative humidity specified as 50%, no significant error will be introduced if no correction for humidity is made. Some ionization chambers have walls made of A-150 plastic or graphite-coated nylon. Such materials are hygroscopic, and the deformations produced by changes in water content can cause substantial changes in sensitivity.

Ion recombination

The polarizing voltage across an ionization chamber limits the amount of ion recombination which can occur. The extent of this recombination depends upon the design of the chamber and the rate of ion production (i.e. instantaneous dose rate) as well as the applied voltage. Linear accelerators produce radiation in pulses, and hence have a much greater instantaneous dose rate than cobalt units. Scanned beam systems produce even greater instantaneous dose rate, and therefore suffer more from recombination.

Various correction methods to account for ion recombination are available. Equations for calculating the ion recombination correction factor can be found in the various protocols. Because the applied voltage, V , as well as the distance between the electrodes is important in these calculations, it is necessary to check whether this distance corresponds to the manufacturer's specification. The recombination correction can also be assessed experimentally by plotting the reciprocal value of the reading against the reciprocal of the square of the high voltage, or by using the two voltage method, as described in IAEA Report 277 and most dosimetry protocols.

For most ionization chambers the ion recombination factor is close to unity for cobalt units. For pulsed accelerator beams the correction is generally less than a few per cent for most types of ionization chamber. Only for scanned beams corrections up to 20% occur for some types of ionization chambers.

Polarity effect

The polarity effect, defined as the difference in response in making identical irradiations of the chamber with and without reversed polarity, is generally well below 1% for cylindrical chambers in photon beams. For plane-parallel chambers the polarity effect can, however, be larger as discussed by Gerbi and Khan [1987] and is due to the lack of charged particle equilibrium. This effect is particularly important for percentage depth dose and tissue-phantom ratio measurements in the build-up layer. Chamber design and electron contamination of the beam, and thus field size, therefore influence the magnitude of the polarity effect. At depths where electron

equilibrium exists, the positive charge produced by the ejected secondary electrons is compensated by the negative electrons stopping in the collector, thus offsetting the polarity effect.

Cable effect

Irradiation of the cable of an ionization chamber must be avoided as far as possible, as the radiation can induce a charged particle flow across the cable. The passage of these particles into or away from the central core of the cable can modify the signal, and hence lead to inaccuracies in dose determination. This effect will be different in-phantom from irradiation of the cable in air. At high photon beam energies the effect will be large in air particularly for thin cables because the central core of the cable will then be in the build-up region. For such a situation a considerable polarity effect will also occur. If by inappropriate design or construction of the ionization chamber, air cavities are present in the stem or in the connection between the stem and the cable, an additional signal will be measured if the field includes this part of the stem or cable.

The cable effect can be measured in two ways:

- (1) The chamber is positioned with its axis parallel or perpendicular to the long axis in the centre of an extremely elongated field thus including or avoiding the irradiation of the stem and cable.
- (2) The ionization chamber is positioned in the centre of an extremely elongated field with various lengths of the cable shielded with blocks.

In well-designed ionization chambers the cable effect during measurements in a large water phantom can be negligible. For measurements in air, i.e. in a mini-phantom, the cable effect can often no longer be neglected and should be determined experimentally. The cable effect varies also with field size and will therefore directly influence the determination of quantities such as $O_D(c)$ and $V(z,s)$.

Stem effect

Ionization chambers should have insignificant stem effects caused by scatter from the stem modifying the dose within the collecting volume. The magnitude of this effect can be determined by investigating the effect of introducing a dummy stem opposite to the actual stem. In well-designed chambers the stem effect at megavoltage photon energies can be less than 0.2% of the reading and can be effectively ignored.

8.2.4. Quality Control of Ionization Chambers.

In order to ensure high accuracy in the absolute dose determination, it is essential to confirm that the sensitivity of the field instruments remains constant. The factors which may be expected to vary the calibration factor of a field instrument include changes in the calibration factor of the local reference chamber and damage of or maintenance work to the field instrument which may be expected to cause a sensitivity change, such as thimble cap replacement.

In the absence of these, the derived calibration factor is expected to remain constant with time for a correctly functioning instrument. Random uncertainties in the calibration process will cause some variation from one calibration to the next. It is therefore useful to adopt a calibration window (e.g. $\pm 0.5\%$) within which no change in calibration is deemed to have occurred.

The sensitivity of the ionisation chamber must remain constant from one calibration event to the next. The consistency may be checked by subjecting the chamber to a known radiation field. One way to achieve this is by using a fixed arrangement of radioactive sources (e.g. ^{90}Sr , ^{137}Cs or ^{60}Co -sources) which are held in precise geometry relative to the chamber and for which the decay is accurately known. Any substantial variation in corrected check source reading should always be followed by a new calibration.

Comparison should be made with consistency check readings taken at the time of calibration of the instrument and immediately afterwards. As for calibrations, it is necessary to adopt window levels (e.g. $\pm 1\%$) within which the instrument is deemed to be operating satisfactorily. Deviation from the expected check value necessitates investigation, and if confirmed suggests the need for recalibration of the dosimeter. It must be noted that the local reference chamber should also be subjected to consistency checks to ensure that its sensitivity remains constant. More details about stability checks can be found in IAEA Technical Report No. 374 (IAEA, 1994).

It is desirable that radiographs of each chamber are taken to ensure that no distortion has occurred to the internal electrode structure. Such distortion may generate a non-isotropic response, and may vary the sensitivity in pulsed radiation fields. Two orthogonal radiographs are necessary to reveal any misalignment of the central electrode. Also unwanted air cavities, causing a cable effect, may be discovered in this way.

For each ionization chamber it is necessary to maintain a record of entries detailing all maintenance work, consistency check results and calibration values for

that instrument. Variations in calibration factors or consistency check results point to problems with the instrument or its method of use or calibration. Inspection of this record prior to use will help to ensure that recalibration of the instrument is always carried out after any maintenance work which is likely to effect the instrument performance.

8.3. DOSIMETRY PHANTOMS

8.3.1. Large Water Phantoms

All modern dosimetry protocols specify water as the reference medium in which absorbed dose is measured. A suitable phantom for absolute dose measurements will allow the ionization chamber to be held precisely at the reference position with the beam incident either vertically onto the water surface or laterally through a thin plastic window. The ionization chamber should be held in a thin PMMA water-tight sheath as described in most dosimetry protocols. The phantom should be transparent to allow visualisation of set-up, and carry reference marks to allow accurate beam alignment. The size of the phantom should be such that it extends at least 5 cm beyond the edges of the radiation field, and at least 5 cm in depth beyond the point of measurement. A suitable phantom is the IAEA reference water phantom (IAEA, 1987).

For measurement of doses relative to the dose at the reference point, e.g. measurements of $P(z,s,f)$ and $T(z,s)$, access to a scanning system which allows remote (or automatic) positioning of the ionization chamber is advantageous. Where such a system is used, the accuracy of positioning should be better than ± 0.5 mm. Such systems tend to use large water tanks, allowing measurement of large field sizes. Many manufacturers of scanning water phantoms use the carriage of their system to move a densitometer to read films. Modern PC-based water phantoms allow not only control of ionization chamber position but permit also on-line and off-line data handling. Some of these systems are able to perform also integrated measurements if a higher accuracy is needed. When such a scanning water phantom is taken into use, its mechanical accuracy and precision must be verified. Data obtained with the scanning system should be compared with static measurements to verify that the scanning system, and the data processing works properly.

8.3.2. Mini-Phantoms

Mini-phantoms are required for a number of measurements including output ratios "in air" and volume scatter ratios. For that purpose a small ionization chamber is positioned at a depth of 10 cm inside a mini-phantom. Such a mini-phantom may

have a square shape of cross section perpendicular to its axis, or consists of a beam-coaxial narrow cylindrical polystyrene or PMMA phantom (rod phantom) as proposed by van Gasteren et al. (1991). The diameter of the rod phantom should be sufficient to provide lateral electron equilibrium but should be as small as possible to avoid side scatter. It is recommended to use a diameter of at least 4 cm, which is suitable for most photon beams in clinical use. It should be noted that in most equations ratios of quantities are required which are less sensitive to possible deviations of lateral electron equilibrium than the single measurements.

Small cylindrical ionization chambers, for instance 0.1 cm^3 chambers as used with a scanning water phantom, have been used to make measurements in a mini-phantom. It is important to correct the chamber readings for the cable effect, particularly if thin cables are employed and high-energy X-ray beams are studied (see section 8.2.3). It is therefore recommended to have the shortest possible length of cable inside the field.

Figure 8-1 shows some examples of mini-phantoms used for the determination of output ratios and volume scatter ratios. For some types of measurements, e.g. the determination of the linear attenuation coefficient, it is useful to vary the depth of the chamber inside the mini-phantom. For that purpose some parts of the material can be removed from or added to the solid mini-phantom, while the water layer can be varied in height in a water-filled mini-phantom.

Build-up caps of high-Z materials, e.g. brass iron or lead, are sometimes used in combination with ionisation chambers to measure output ratios in air. For field sizes varying between 5×5 and $30 \times 30 \text{ cm}^2$ the difference observed between output factors obtained using ionisation chambers in combination with these build-up caps or a PMMA mini-phantom are less than 0.5%. For more extreme field sizes the differences are larger, which might be due to the higher density of the build-up cap materials thus causing a larger scatter than that in the mini-phantom. It is therefore recommended to use the polystyrene, PMMA or water-filled mini-phantom for measurements "in air".

8.3.3. Other Phantoms

For a number of dose measurements, such as measurements in the "build-up region or day-to-day calibration check measurements, the use of a large water phantom is not strictly necessary and use may alternatively be made of a solid phantom. A wide variety of solid materials may be used, including "solid water", polystyrene or PMMA. Slabs of solid phantom material with dimensions of about 30

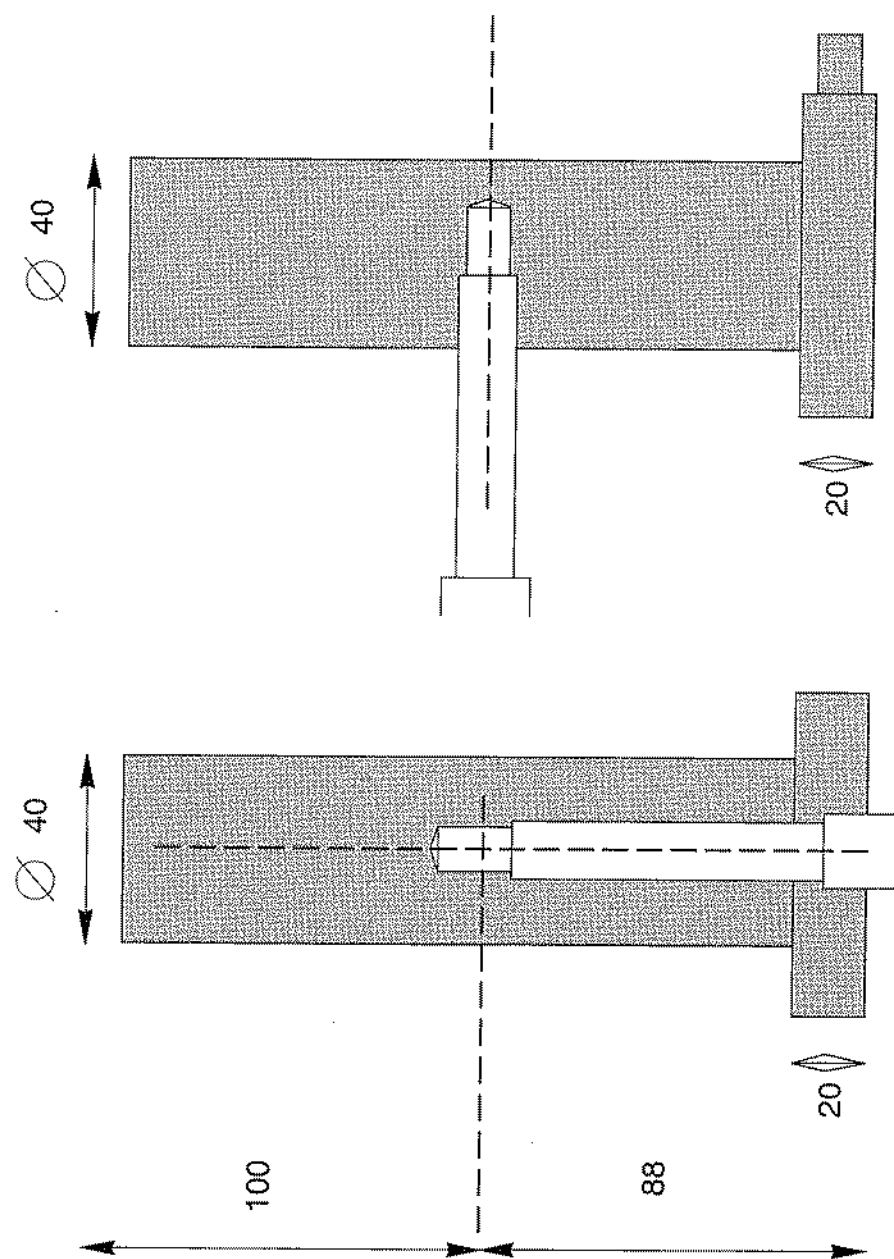


Figure 8-1. Examples of plastic mini-phantoms; left) a small ionization chamber is positioned with its axis parallel to the beam axis; right) a small ionization chamber is inserted perpendicular to the beam axis;

cm x 30 cm and several cm thick are very useful for many types of dose measurement. In one of these slabs a hole should be drilled to accommodate an ionization chamber. A slab with an opening machined to fit a plane-parallel ionization chamber with its entrance window at the level of one surface is useful for making measurements in the build-up region.

When a solid phantom is adopted for use in a specific phantom beam, the dose measured in it must be cross-referenced to the dose measured using a reference water phantom such that a phantom correction factor may be determined. Although such a correction can in principle be determined theoretically, as described in some dosimetry protocols, it is recommended to apply experimental ratios because theory requires not only material dependent parameters but also accelerator specific information related to the head scatter.

8.4. MEASUREMENT PROCEDURES

8.4.1. Dose under Reference Conditions.

The formalism applied for dose determinations under reference conditions in a high-energy photon beam, using the ionization chamber method, is in principle the same in most dosimetry protocols. First the field instrument is calibrated in terms of exposure, air kerma or absorbed dose to water. The calibrated field instrument is then used to determine the absorbed dose to water at a reference point in a water phantom irradiated by the photon beam. In most dosimetry protocols the depth of the reference point is varying according to beam quality. It is recommended in this report to apply only one reference depth of 10 cm for all beam qualities.

8.4.2. Output Ratios and Volume Scatter Ratios

According to the recommendations given in this report, output ratios (O) are defined at 10 cm depth. Output ratios are required for the situation under full scatter conditions in a large water phantom as well as 'in air', i.e. measured in a mini-phantom. In order to measure output ratios most types of ionization chamber are suitable. For large field sizes a Farmer-type chamber, which is also often used for the dose determination under reference conditions, is appropriate. For small field sizes a chamber with a shorter effective length is necessary.

Output ratio measurements under full scatter conditions, O_R , should be performed in a large water phantom, for instance in the water tank of the scanning phantom using the small ionization chamber of that system. Output ratios may also be determined in polystyrene or water equivalent plastic. Small differences with

water measurements for the same field sizes might occur due to differences in electron density between the various materials. Although in principle the field size can be scaled for different phantom materials, this is only valid for the phantom scatter component. Differences in head scatter for different field sizes should be taken into account, which makes conversion of data measured in other phantom materials to output ratios obtained in water, rather cumbersome.

Output ratios 'in air', O_0 , should be measured with a mini-phantom as described in Section 8.3.2. These output ratios should not only be available for square fields but also for rectangular fields. Due to the difference in position of the inner and outer collimator pairs (see figure 3-3), output ratios will depend on the choice of the collimator which shapes the field, resulting in the collimator-exchange effect. As discussed in section 5.6 use can be made of an expression proposed by Vadash and Bjärngard (1993) to find the equivalent square collimator setting O_R for the rectangular setting (L, W):

$$c_e = (A + 1) LW / (AL + W)$$

where A is a parameter specific for each beam. A values between 1.4 and 1.8 have been reported (Vadash and Bjärngard 1993, Georg and Dutreix 1997). They should be determined experimentally for each accelerator. For that purpose a set of output factor values should be measured keeping one collimator setting fixed and varying the other one.

Volume scatter ratios, V , are dose ratios measured under full scatter conditions and in a mini-phantom. Both sets of data should preferably be measured with the same dosimeter, in the same measurement session to avoid instabilities in the accelerator output or measurement system.

It should be pointed out that according to their definition both V and O_0 rely upon absorbed dose measured in the mini-phantom. Dosimetry protocols do, however, not provide sufficient data to derive the absorbed dose to water from an ionization chamber reading obtained in a mini-phantom. In practice, therefore, the ratio of two readings is assumed to be equal to the ratio of the two dose values. For O_0 this assumption will generally not introduce large uncertainties. The value of V will, however, depend on the measurement technique, i.e. the type of detector applied for these measurements. For instance the energy dependence of the detector will influence the resulting V value. In all equations for the computations of monitor units, only ratios of V values are applied in which these differences will largely cancel.

8.4.3. Tissue-Phantom Ratio and Percentage Depth Dose

There are various ways in which percentage depth dose data in a water equivalent phantom, P , can be collected. A few careful point measurements for various field sizes along the central axis of the beam, in combination with a fit or formula, is generally sufficient to describe adequately depth dose curves. Both water phantoms and solid phantoms can be used for this purpose. However, these data can most easily be obtained using a remotely controlled scanning water phantom. A number of commercial systems is available using small ionization chambers, or in some cases, specially designed diodes. Although these systems differ in their mechanical aspects, detection electronics and data acquisition and analysis software, the resulting depth dose data should in principle be sufficiently accurate (if an adequate acceptance testing procedure has been performed). Small ionization chambers, e.g. with a sensitive volume of about 0.1 cm^3 , are generally provided with these systems. A second chamber is often employed as a reference chamber to correct for dose rate fluctuations of the accelerator.

These small ionization chambers generally cannot be used in the build-up region. For that kind of measurements it is necessary to use a parallel-plate detector with a thin wall. Again, integrating point measurements can be made as well as relative dose rate measurements using a scanning device if the depth is accurately known.

Tissue-phantom ratios, T , are more tedious to measure than depth dose curves because the thickness of phantom material has to be varied for each measurement. Some of the commercial water phantom systems are capable of remotely controlling the water layer above the detector, thus providing experimental T values in a relatively simple way. An alternative method is to calculate T values from percentage depth dose values (see Appendix 1). These calculated T values should be verified with measurements at some depths for some commonly employed field sizes.

If solid phantoms are applied for measurements of tissue-phantom ratios or percentage depth doses, conversion of the measured data to those in water is required. Such a conversion should not only be based on the difference in specific density and atomic composition of the two phantom materials but also include possible depth-dependent reading to dose conversion factors of the applied dosimeter. Because such a conversion will always introduce additional uncertainties, it is recommended to apply water-based P or T values in all monitor unit computations.

8.4.4 Wedge Factors

The wedge factor is defined in this report at 10 cm depth in a water phantom or in a mini-phantom along the central axis of the beam. Wedge factor measurements can thus be performed with a Farmer-type chamber, either directly in a water phantom or at an equivalent depth in a solid phantom or in a mini-phantom. Wedge factors vary with field size due to a change of the primary photon fluence, and therefore have to be determined for various field sizes at 10 cm depth. The variation of the wedge factor with depth for a particular field size should be incorporated as a set of different tissue-phantom ratios or percentage depth dose data for wedged fields compared with open fields. The wedge factor decreases slightly with increasing source-skin distance. Wedge factors should therefore be measured at a few clinically relevant SSDs.

Wedge factors should be measured with the axis of the ionization chamber in the direction perpendicular to the wedge profile. The measurements should be made twice with the wedge orientation rotated 180 degrees. The average of the two measurements should be taken as the correct value. If the difference between the two measurements is larger than several percent, the position of the ionization chamber or the wedge should be checked.

8.4.5 Tray Factors and Shielding Block Factors

Shielding blocks are generally placed on a plastic or aluminium plate which is positioned under the gantry of an accelerator or cobalt machine. The tray attenuates the primary beam and will therefore reduce the dose at the dose specification point compared with the situation without tray, for the same number of monitor units. This tray factor, k_t , should be measured by placing an ionization chamber at the reference depth of 10 cm either in a water phantom or in a mini-phantom. Because k_t depends slightly on field size due to photons scattered in the shadow tray, measurements should be performed for a few collimator openings.

The presence of a tray not only attenuates the primary photon beam but produces scattered photons and secondary electrons. For that reason the tray will also modify the percentage depth dose or tissue-phantom ratio data, valid for the open beam, particularly at short distances (see figure 4-1). A tray with holes is often applied to position blocks having a specific divergence. An average tray factor for such a tray can be obtained by taking the average ratio of profile measurements with and without the tray. These measurements can be performed with a scanning water phantom or using film dosimetry. It should be noted that if such a tray is used during

in vivo dose measurements, the position of the dosimeter becomes very critical and may cause an increase in the standard deviation in repeated measurements.

The shielding block factors, k_b and $k_{b,o}$, which correct for the modification of the dose by positioning shielding blocks on the tray, have to be measured at 10 cm depth in a water phantom or in a mini-phantom, respectively. Because k_b and $k_{b,o}$ depend on the distance of the point of measurement to the edge of the shielded area, it is recommended to measure these factors with a small ionization chamber. In principle k_b and $k_{b,o}$ should be determined for each block configuration, but approximate values can be estimated from a limited number of measurements (see section 6.3).

8.5. NUMBER OF MEASUREMENTS REQUIRED

8.5.1. Measurements Required for Manual Calculations

In most radiotherapy departments not all patients are treated with complicated techniques requiring a computer calculation of the number of monitor units. Often anterior-posterior fields are applied, or sometimes single fields, for which only a simple manual calculation of the number of monitor units is needed. For these situations each department requires at least sets of tables of output ratios and tissue-phantom ratio or percentage depth dose data. In addition tray factors and other correction factors for accessories applied during patient treatment are required. For more complicated treatment techniques additional data such as wedge factors, the collimator exchange effect and off-axis ratios are needed.

When a new accelerator is accepted, the beam parameters are fixed and these data have to be collected before patient treatment starts, i.e. during the commissioning of the equipment. In that period ample time should be available for physics measurements. Once equipment is installed it might be profitable to measure a large amount of data instead of measuring only a limited set of data and applying interpolation or fitting procedures. Use can be made of existing information for a certain type of equipment. If, for instance, the quality index of a photon beam is known, existing sets of T or P values can be used which have to be checked with a limited number of measured data. Output ratios and volume scatter ratios should always be measured for each accelerator because they depend on the dimensions, type of material and geometry of the flattening filter and monitor chamber, which might differ for each accelerator. Ratios of V values, as required in many monitor unit (U) computations, are only dependent on the quality index and can be compared with existing information (e.g. see Storchi and van Gasteren, 1996).

Most numerical data can be obtained using integrating measurements at specific predetermined points. Alternatively, the measurements required for the computer planning system may also be used for constructing tables for manual calculations. This can sometimes be done within the software of the scanning water phantom system. Otherwise the data required can be obtained with a digitizer from the plots produced on the water phantom system.

It should be noted that quality control and maintenance programmes should be aimed at keeping the characteristics of the accelerator identical to the situation during the acceptance testing and commissioning. If after major repair or for other reasons the beam properties might be changed, a large amount of data has to be remeasured or at least checked.

8.5.2. Measurements Required for a Computer Treatment Planning System

Treatment planning systems require input data to their beam models for generating isodose curves and sometimes for the calculation of monitor units. These data are generally obtained with a scanning water phantom system. Tissue-phantom ratios or percentage depth doses, beam profiles and output ratios are needed. The amount and type of input data vary considerably between the various planning systems. Some planning systems require separate data for wedged beams while other systems generate wedged data using an accurate description of the wedge dimensions and position coupled with beam attenuation characteristics of the wedge material. In both types of planning systems, the field size dependence of the wedge factor has to be measured. Asymmetric beam data are required for some planning systems while additional build-up measurements are needed for a more accurate description of the dose distribution in the first centimeters in a phantom or patient. In addition to the basic beam data, a number of measurements has to be performed to test algorithms of the planning system for specific geometries. For instance the shielding block algorithm and the dose calculation for wedged beams should always be tested for a number of clinically relevant situations. As far as the calculation of monitor units is concerned, these tests should be specific for the algorithm applied in a particular institution or treatment planning system. It is important to check combinations of algorithms for a range of situations. For example, the combination of shielding blocks and wedges might introduce unexpected errors in the calculation of monitor units. Due to the wide variation both in monitor unit calculation algorithms and in clinical situations, it is difficult to indicate how much effort should be spent by each institution in testing algorithms for monitor unit calculations. Factors effecting these calculations and related quality assurance programmes have been discussed by Van Dyk et al., (1993) and an AAPM task group (Kutcher et al., 1994).

9. QUALITY ASSURANCE

9.1. INTRODUCTION

There are many steps involved in a radiotherapeutic procedure, i.e. the process starting with the delineation of a target volume in a patient to the actual delivery of a prescribed dose to that target volume. All steps are important in determining the overall quality of treatment. However, international and national recommendations for **quality assurance**, QA, are only generally available for the performance tests of megavoltage therapy units and for the determination of dose according to dosimetry protocols. There is some information in the literature on the quality control of simulators, CT scanners and treatment planning systems (e.g. Brahme et al., 1988, Van Dyk et al., 1993, and Kutcher et al., 1994), but most institutions have developed their own QA programmes in the absence of definitive guidance. QA in relation to the calculation of monitor units is even less well developed, with little information being available on the various approaches which may be adopted.

We will start this chapter with a brief review of the various QA programmes available on the various aspects of the radiotherapeutic procedure. The main purpose of this chapter is, however, to discuss QA applied to the calculation of U or treatment time required to deliver a prescribed dose to a specified point in the target volume. We assume that this point is chosen in agreement with the recommendations given in ICRU Reports 29 and 50 (ICRU, 1976, 1993). If for specific reasons the dose is not prescribed at the ICRU reference point but in another way, e.g. the mean dose to the target volume, then knowledge of the ratio of these two dose values should be available. The dose at the ICRU reference point may be related directly to the number of monitor units or irradiation time. This relationship has been discussed extensively in chapters 5 to 7. Methods now have to be formulated to check the results of these calculations in an independent way.

Even with a good QA programme covering the irradiation equipment and the U/irradiation time calculation, deviations in the output of the treatment machine or errors in the treatment procedure may still occur, causing discrepancies between the prescribed and actual dose delivered to the patient. These deviations may be traced using phantom measurements, either under reference conditions or with an anthropomorphic phantom under some clinically appropriate circumstances. It is recommended that these types of measurements are performed by an independent observer, i.e. by means of a quality audit. The role of quality audits and participation in dosimetry intercomparisons will be discussed in this chapter. The use of *in vivo*

dosimetry as an even more complete check of the whole dosimetry procedure for individual patients will be elucidated as well. Finally a very important aspect of a QA programme, documentation and record-keeping, will be discussed.

The exact meaning of concepts such as quality assurance and quality control are not the same in all reports. In this manual we will adopt the definitions of quality assurance, quality control and quality audit as given in IAEA Report TRS 277. These definitions are:

quality assurance for radiation dosimetry consists of the formulation of those procedures that must be taken in order to enable accurate and consistent dose measurements to be carried out;

quality control is the process by which the actual performance is measured and compared with performance standards, and the actions necessary to maintain or regain conformance with the standard identified;

quality audit (assessment) is an independent review of the quality assurance programme and the results achieved by it.

9.2. EXTERNAL DOSIMETRY CHECKS AND QUALITY AUDITS

A quality control, QC, programme for the calculation of U (or treatment time) starts with the assumption that the actual output of the treatment machine is correct within certain specified limits. In order to guarantee such a situation, each radiotherapy department needs a corresponding QC programme for megavoltage therapy units. Even when an appropriate and practical QC protocol is followed in a particular institution, deviations of the "true" value of the dose at the reference point may occur due to various reasons. It is therefore recommended, and in some countries it is mandatory, that an independent person, visiting team or organization checks the output in a completely independent way. Dosimetry quality audits or external dosimetry checks should not be restricted solely to dose measurements under reference conditions but should be extended to irradiations of anthropomorphic phantoms under more clinically relevant conditions. Patient-related errors in the dose calculation procedure, for example in the conversion of reference conditions to isocentric techniques, can be discovered in this way. Other factors important in determining U such as wedges and blocks, can be incorporated in such an audit programme. Other types of errors, such as those related to field size, might not influence the dose at the ICRU point very much but are very important for the success of the treatment. Errors of these type can also be traced with external dosimetry checks by incorporating measurements at off axis points.

Dosimetry quality audits or external dosimetry checks might reveal serious discrepancies between the stated dose and the dose measured during audit. Even when the deviation between stated and measured dose is not large, quality audit is a useful tool to confirm the correctness of a newly established procedure or to indicate when a number of small deviations which may cancel in the final dose assessment for one particular type of treatment, may add together in others to generate a significant deviation.

9.2.1. Results of External Dosimetry Checks and Quality Audits

External dosimetry checks can be performed in relatively simple ways, such as by inviting, informally, a colleague from a neighbouring department to check the output. This can be carried out on a mutual basis. Such a system has been introduced in Great Britain, and has been formalized into the development of a Quality Audit arrangement based upon geographical location, with each Audit Group consisting of a number of radiotherapy centres. In other countries external dosimetry checks by visiting teams have been established in which institutions participate on a voluntary base. It can be expected that in future, dosimetry quality audits will replace these checks as the implementation of ISO 9000 increases. The obligation to undergo a check or audit is extremely important for departments which participate in clinical trials and which want to combine their results.

A number of external dosimetry checks has been conducted over the last two decades. Table 9.1. summarizes the results of various dosimetry intercomparisons under reference conditions for ^{60}Co gamma-ray beams and megavoltage X-ray beams. In most dosimetry comparisons the dose stated by the institution is expressed as the ratio of that value and the dose measured by the visiting team or organization. These ratios are expressed as mean, standard deviation and maximum deviation from unity, where the mean allows assessment of systematic differences (e.g. between the dosimetry protocol used by the various institutions to that of the visiting team), and the standard deviation and maximum deviation are useful indicators of performance across the group of institutions.

From table 9.1. it can be seen that the standard deviation for the X-ray beams is larger than the standard deviation for the ^{60}Co gamma-ray beams. The results for ^{60}Co are generally good except for a few institutions as for instance observed during the study of Johansson et al., (1986) and Thwaites et al., (1992). The latter intercomparison revealed an error of about 25% in the output determination of one ^{60}Co gamma-ray unit.

TABLE 9.1. RESULTS OF DOSIMETRY INTERCOMPARISONS UNDER REFERENCE CONDITIONS.

The dose is determined with an ionization chamber. The number of beams, the mean, one standard deviation (SD) of the ratios of stated and measured dose values, and the largest difference observed between the ratios (delta) are given.

Reference	Year	Region	Beam quality	Number	Mean	SD	Delta
Svensson	1971	Scandinavia	^{60}Co	-	0.996	0.010	0.03
			x-rays	-	0.994	0.059	0.23
Shalek et al.	1976	USA	$^{60}\text{Co}/\text{x-rays}/\text{e}^-$	352	0.995	0.034	0.30
Samulski et al.	1981	USA	$^{60}\text{Co}/\text{x-rays}/\text{e}^-$	254	1.003	0.027	-
			^{60}Co	22	0.999	0.014	0.05
Johansson et al.	1982	Scandinavia	x-rays	50	0.983	0.023	0.10
			^{60}Co	59	0.999	0.019	0.10
Johansson et al.	1986	Europe	x-rays	16	0.976	0.033	0.14
			^{60}Co	11	1.006	0.006	0.02
Wittkämper et al.	1987	The Netherlands	x-rays	40	0.992	0.020	0.10
			^{60}Co	61	0.998	0.014	0.08
Thwaites et al.	1992	UK	x-rays	100	0.997	0.015	0.10
			^{60}Co	13	0.999	0.010	0.04
Hoornaert et al.	1993	Belgium	x-rays	21	0.994	0.023	0.08

Several external dosimetry checks have been conducted using a phantom with anatomical structures under clinical conditions. In table 9.2., data have been summarized for a few recent intercomparisons. Dose values are either measured with an ionization chamber or with TLD. The standard deviation in the ratio of stated and

measured dose values at the centre of the target volume varies between 1.5% and 3.4%. The latter value reduces to 2.7% for a homogeneous situation. At other points in the target volume the uncertainty increases, partly because of additional uncertainty in the TLD measurements. The reproducibility of the ionization chamber measurements is generally better than 1%, whereas the precision of most TLD measurements is approximately 2%.

TABLE 9.2. RESULTS OF DOSIMETRY INTERCOMPARISONS OF PHOTON BEAMS UNDER CLINICAL CONDITIONS

The number of beams, the mean value, the standard deviation (SD) and maximum deviation (delta value) are given of the ratios of stated and measured dose values. The first result gives the data at the centre of the target volume whereas the second line applies to other points in the target volume.

Reference	Year	Region	Tumour site	N ^a	Mean	SD	Delta
Johansson et al.	1987	Europe	tonsil	15	0.96	0.023	0.06
				17	0.98	0.05	0.18
Witkämper et al.	1987	The Netherlands	prostate	36	0.985	0.015	0.06
				207	1.004	0.052	0.30
van Bree et al.	1991	The Netherlands	breast	21	1.021	0.028	0.11
				462	1.04		0.20
Thwaites et al.	1992	Great Britain	mediastinum	62	0.989	0.034	0.24
				61	0.992	0.035	0.25

Independent verification of output calibration by an on-site allows immediate identification of variations in procedures. A similar check can, however, also be performed by postal dosimetry. Several postal dosimetry services are offered by various organizations such as the IAEA/WHO (Svensson et al., 1990) and the RPC (Hanson et al., 1991) and are represented in table 9.3. The IAEA recently modified the cobalt-60 TLD system which has been used for many years, so that it is applicable to high-energy photon and electron beams. The system uses a hollow rod to hold the TLD which can be placed at fixed depths in a water phantom in the user's beam. The

IAEA/WHO postal dosimetry service includes several hundred hospitals per year. In some parts of the world, this postal service is instead carried out by National Secondary Standard Dosimetry Laboratories, which are connected to IAEA/WHO SSDL network. The IAEA dosimetry laboratory in Vienna then functions as a reference centre, by exchanging TLD dosimeters with the National SSDL to assure the traceability to BIPM.

TABLE 9.3. RESULTS OF MAILED TLD DOSIMETRY INTERCOMPARISONS OF PHOTON BEAMS UNDER REFERENCE CONDITIONS.

The mean and one standard deviation (SD) of the ratios of stated and measured dose values are given.

Reference	Time period	Organization ^d	Number	Mean	SD
Nette et al. 1994	1969-1994	IAEA	2403 ^a	1.007	0.091
Hanson et al. 1991	1984-1991	RPC	8895 ^b	1.000	0.024
Hansson et al. 1993	1987-1992	EORTC	357	1.007	0.040
Dutreix et al. 1994a	1991-1993	EC	125	0.970	0.095
			119 ^c	0.985	0.025
Izewska et al. 1995	1994-1995	EROPAQ	129	1.001	0.074
			122 ^c	0.998	0.027

^a excluding deviations larger than 30%

^b excluding deviations larger than 25%

^c excluding deviations larger than 12%

^d IAEA : International Atomic Energy Agency

RPC : Radiological Physics Center

EORTC : European Organization for Research and Treatment of Cancer

EC : European Commission ("Europe Against Cancer")

EROPAQ : European Radiation Oncology Programme for Assurance of treatment

The organisation and role of the IAEA/WHO network of SSDLs, and further development of the system have been described elsewhere (Svensson et al., 1990). The results of the RPC-mailed TLD system showed that for 9000 photon beams, 95% of the measurements were within their 5% criterion. Postal dosimetry systems require procedures to investigate deviations in excess of predetermined criteria in order to identify causes and propose corrective actions.

External dosimetry checks and quality audits should be performed on a regular basis. The frequency depends on many factors, such as the local QA programme and resources available for dosimetry checks. A minimum frequency for dosimetry checks is recommended as every five years. After major repair to the unit or to the dosimetry equipment, it is recommended that an external dosimetry check be performed before patient treatment recommences.

9.2.2. Origin of Deviations Observed During External Dosimetry Checks and Quality Audits

External dosimetry checks mainly verify the results of a complex procedure such as the dose measurement under reference conditions. If any serious discrepancy is observed, it should be investigated immediately to determine the cause and probable impact. It is recommended therefore that the report of a dosimetry check is as extensive as possible with sufficient details as to the measurement conditions and dose derivation procedure to allow the visited centre to explain the observed deviations. Preferably, these data should be available during the actual visit to allow immediate clarification and assessment, if necessary. It is, in any event, always a fruitful and stimulating experience to discuss dosimetry procedures in a department with an independent observer.

There are many reasons for differences between stated and measured doses as discussed by the various authors mentioned in tables 9.1. to 9.3. Firstly, arithmetic errors or incorrect interpretation of a dosimetry protocol have been discovered. Interpolation between data presented in tables or the reading of graphs has introduced errors. A very common cause of discrepancy results from the replacement of a procedure or the use of a new data set or protocol without modifying the data at all places in the clinic. During dosimetry intercomparisons it has been observed, for example, that good agreement may exist between dose values determined by the local physicist and the visiting team, but that discrepancies have occurred between stated and measured dose values because old tables were applied in the clinic. Output fluctuations have also contributed to observed discrepancies. In the study of Hoornaert et al. (1993) about 1/3 of the standard deviation could be ascribed to

accelerator fluctuations. The introduction of a common code of practice or dosimetry protocol has resulted in a noticeable reduction in standard deviation and maximum deviation (delta value) in some of the studies mentioned in tables 9.1. and 9.2.

Ionization chamber dosimetry requires the application of a number of conversion and correction factors as discussed in Section 8.2. Occasionally, correction factors may be unjustly ignored or underestimated. An error in the recombination correction was one of the main reasons for a large standard deviation and delta value for high-energy electron beams in the European study of Johansson and colleagues (1986). In addition, a number of practical problems with temperature or pressure measurements, in the determination of the target-phantom distance or in the accurate positioning of the chamber in the phantom have been observed.

Additional causes for deviations have been identified during the dosimetry intercomparisons under clinical conditions. For the intercomparisons presented in table 9.2., the dose was calculated using computer planning systems. It was observed that in some of these planning systems beam data were applied that did not correspond to the actual beam data. A QC programme of the accelerator should therefore always be accompanied by an additional QC programme of the planning system.

9.3 QUALITY CONTROL OF MONITOR UNIT CALCULATIONS

9.3.1. General Principles

In order to design a QC programme covering U calculation for a particular department, it is useful to analyse the various steps in the calculation and delivery of a dose related to a specific number U to a patient. This process is illustrated in figure 9.1. Firstly the relevant beam data, for the irradiation device have to be measured and implemented in the U calculation system. This can be:

- a computerized treatment planning system (TPS),
- a U calculation programme run on a separate computer applying tables stored in that system,
- a hand calculation which uses tables of beam data.

By using patient data, obtained from simulation, CT-scans or other diagnostic sources of information, in combination with these beam data, U can be calculated for a specific prescribed dose. The most important data relevant for this calculation are then transcribed onto the treatment chart, which is to be used as input for the

treatment machine. The input into the treatment machine can be done either manually at each treatment session or by means of a patient file as for a modern computer controlled accelerator having a verification system.

Quality control of beam data is generally well organized in most radiotherapy departments and a number of national and international QC programmes for Co-60 units or medical accelerators are readily available. Guidelines for a QA programme to guarantee correct patient data are also available for simulators and CT scanners. QC programmes for mechanical aids and image quality should be available in each department covering other diagnostic equipment and tools applied for measuring patient contours. In addition to these QC programmes, the calculation of monitor units for individual patients needs to be subjected to careful and systematic control to ensure that the results are accurate. Finally the actual dose delivered to the patient should be compared with the prescribed dose which is related directly to the calculated number of monitor units.

In principle, three sources of error may exist which influence the accuracy of the U calculation. First the beam data applied in the various calculation systems should correspond with the actual beam data. Differences may occur even if both the equipment and the U calculation procedure are subject to a regular QC programme. These discrepancies were in fact the major source of error in an intercomparison of measured and calculated dose distributions for the treatment of breast cancer with tangential fields (van Bree et al., 1991). The second type of error that may occur, relates to systematic errors in the dose calculation method. For example, a simple "hand" calculation will in general not take into account the full 3D geometry and will therefore have limitations to its accuracy. Also undetected deficiencies in the calculation might occur if the dose calculation algorithm is not fully tested under all clinical circumstances. This situation may typically occur when a new irradiation technique is introduced in the clinic. The third and probably most important type of error, relates to human mistakes. These may be due to various reasons such as misreading a number from a graph or a table, arithmetical errors or transcription errors.

The relation between prescribed dose and monitor setting should therefore start with a check of the calculation of the correct number of monitor units for each patient. Errors in the transcription of the results of these calculations onto the treatment chart and into the verification system of the treatment unit require additional steps, and should be checked by an independent second operator. Finally the actual patient treatment may differ from that prescribed on the treatment chart. In

Fig. 9-1 : Schematic presentation of the various steps involved in the calculation of the number of monitor units and the delivery of the dose related to that number during a specific patient treatment.

order to trace these errors in vivo dosimetry is a useful method of QC. These various QC programmes will be discussed in the next sections.

9.3.2. Independent QC Programmes

The first step of any programme established to check the calculation of monitor units is to check the patient data. It is recommended that these patient-specific input data, whether derived from the physician's prescription, the simulation and other preparatory data-collection, or from the graphic treatment planning process, i.e. using CT scans, be verified by an independent second operator. This should take the form of a review of all such data, checking against the original documents or images. Both operators need to sign and date the document that lists these parameters.

The next step is to check the machine related input data for each patient treatment. These data concern a detailed description of the treatment technique (e.g. beam energy, field sizes, gantry rotation, collimator rotation, wedge angle) as well as the detailed dose/fractionation schedule including dose values at the dose specification point, the dose maximum and their values integrated over the whole treatment period. These data should be checked independently by a second person and a check form listing these data should be signed by both persons.

If the input data has been checked for accuracy, then application of a computer to calculate the U setting from these data may be expected to always yield the same result. Under these circumstances, it is doubtful if a simple independent recalculation using the same computer system is worth while. Verification of the calculation should be performed independently from the original calculator. Such independence may be achieved, for example, by using a separate computer system which uses separate algorithms of reduced complexity (and accuracy). In this way, errors of greater than $\pm 3\%$ may be identified. When no second computer is available, a check may be performed manually using look-up tables. The data entered into the treatment chart or entered into the verification system may be used as the starting point for check calculations. Additional patient data might be required from the "setting up instructions" provided at the simulation. If the QC calculation is simple and fast, it can be performed by the radiographers at the treatment units. Such a procedure has the additional benefit that it gives them an increased confidence in and a better understanding of the treatment which they are delivering. Otherwise the checks should be performed by physicists. Such a procedure might reveal a number of errors such as differences in SSD, wrong use of depth dose or output tables or transcription errors. Serious errors of 5% or more were detected in this way in about 0.5%

(Williams et al., 1991) and 1.3% (Calandrino et al., 1993) of the total number of plans checked. If the QC is performed in the first week of the treatment, any error can easily be compensated for subsequently.

This kind of QC of U calculation is greatly facilitated by the presence of printouts from either the treatment planning system or from the separate computerized U calculation programme. If the calculation is done manually, a form with the data relevant for the calculation (which can sometimes be the treatment chart) should be available. Computer printouts intended for this function only may not necessarily be considered permanent records but rather worksheets. If the printout includes output ratios, tissue-phantom ratios or percentage depth doses, equivalent square fields, tray factors and wedge factors, an experienced reviewer will find the tedium of these routine checks greatly reduced. In more complicated geometries, using sophisticated U calculation algorithms, printout of the total dose contributed by the beam to the prescription point, followed by a breakdown of its components to include output ratios "in air", volume scatter ratios and correction factors for the presence of blocks and anatomical heterogeneities, can be of great value to the physicist who invariably must review at least these more complicated calculations. The manufacturer and the user of a treatment planning system should together design worksheet printouts that will suit the working arrangements of the department.

A QC protocol for U or treatment time should not only give a description of the test method, i.e. the formula and data sets to be used in the calculation, but should also include the tolerances (action levels), the actions to be taken, the frequency of checks and the person responsible for the checks. Documentation is an important part of a QC protocol and will be discussed separately.

9.3.3. In Vivo Dosimetry

In any radiotherapy institution where it is already common practice to check the calculation of U (or treatment time) by two persons independently, the additional effort of having a QC protocol for this particular part of the radiotherapy process should be carefully considered. The chance of making undetected errors is reduced by using a treatment planning system that calculates monitor units or by applying a separate computer programme for U computation. Once such programmes have been tested and accepted, the number of mistakes which relate to the wrong use of tables or to arithmetical errors may be reduced considerably. The amount of time and expertise required for a careful verification of the monitor unit calculation can be very substantial, e.g. if off-axis points and heterogeneities are involved or some dynamic conformal therapy scheme is used. In these situations it might be a better strategy to

introduce in vivo dosimetry in a systematic way in a department such that not only errors in the monitor unit calculation can be discovered but also other types of mistakes, for instance related to patient data, patient setup and transfer of data.

The various techniques applied for in vivo dosimetry have been described elsewhere (Mayles et al., 1993 and Van Dam and Marinello, 1994). The most commonly applied method is to place a silicon diode or TLD on the patient during the irradiation and to compare the measured dose with that predicted by calculations. Routine use of this technique has been demonstrated to be effective in identifying discrepancies of a few percent, sufficient for verification purposes. In general, entrance doses are verified but for a more accurate assessment of the target absorbed dose, e.g. when anatomical structures including tissue heterogeneities are of concern, exit doses need to be measured as well. The results of in vivo dosimetry should be treated as part of the patient's permanent record. A number of studies has shown the usefulness of in vivo dosimetry in tracing errors in the actual dose delivered to the patient (e.g. Leunens et al., 1990, Van Dam and Marinello, 1994, Noel et al., 1995 and Essers, 1996).

Table 9.4 shows as, an example, the type of errors in the U computation that can be traced using in vivo dosimetry in combination with an extensive set of phantom measurements (Heukelom et al., 1992). The monitor units for this group of patients were routinely calculated using a commercial treatment planning system. At that time the algorithm did not separate head and phantom scatter which introduced an error up to -1.6% in the calculated output for an SSD of 80 cm instead of the correct output at 100 cm SSD. Also the collimator exchange effect as well as the wedge factor dependence on depth, field size and SSD were not taken into account at that time. The summation of a number of small systematic errors resulted in a value of the dose at the isocentre which was about 7% higher than the dose calculated using the MU programme. Based on these results, improvements in the monitor unit calculation algorithm were implemented in the treatment planning procedure of prostate patients entering in a dose escalation study. In vivo dose measurements have since then been routinely performed on all these types of patients, with adjustment of the number of monitor units if the discrepancy between measured and prescribed dose is larger than 2.5 % (Essers, 1996). Fig. 9.2 shows the results of these measurements over the period of about a year. The unexpected discrepancies starting with patient number 36 were due to a change in the monitor unit calculation programme: the dose normalization was modified from dose maximum of each beam to the dose specification point. Although the modified algorithm was tested for a number of situations, an error in the U calculation for this technique (short SSD for the lateral

fields and the use of transmission blocks) was only noticed after the actual treatment of patients.

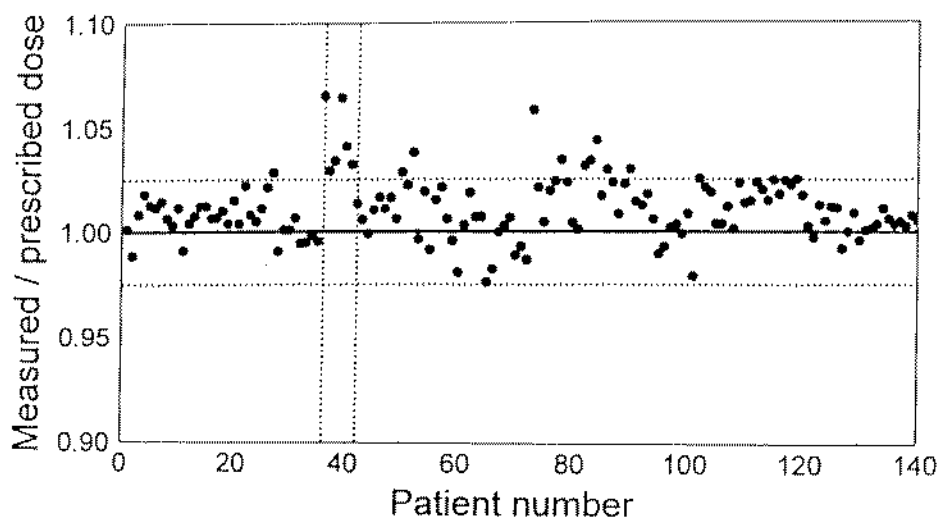


Fig. 9-2 : The ratio of measured and prescribed dose at the isocentre for patients treated according to the protocol for conformal therapy of prostate tumours. The black squares are the measured data before a correction is applied which is indicated by the arrows. The dashed line is the action level of 2.5%

TABLE 9.4. ERRORS IN THE U COMPUTATION TRACED BY IN VIVO DOSIMETRY IN COMBINATION WITH PHANTOM MEASUREMENTS (from Heukelom et al., 1992).

Maximum deviation between the dose calculated by the 2D-treatment planning system and the dose measured in a phantom, for the isocentric treatment with lateral fields of the pelvis.

Output calculation

Output factor	-1.6%
Collimator exchange effect	-1.0%
Wedge factor (field size)	-2.0%
Wedge factor (SSD)	-1.0%

Calculation of the depth dose distribution

	Entrance point	Isocentre	Exit point
PDD wedged beam	+1.0%	-1.6%	-4.1%
Lack of backscatter	-	-	+1.3%
Overall difference	-4.6%	-7.2%	-8.4%

9.4. DOCUMENTATION AND RECORD-KEEPING

9.4.1. Scope

In this section, we will examine the requirements for record-keeping and archiving of data that are related to U calculations for individual patients. These are not independent of the more general task of documenting results of other quality control programmes. U calculations make use of a set of beam data and assume that the calibration of the unit is perfectly accurate and that the beam data perfectly represent the beam parameters which are measured. It is the purpose of quality control programmes relative to those aspects to guarantee that this assumption can be made, within specified limits of tolerated variations, each time a patient is treated. Thus, there is a great deal of similarity, even redundancy, between record-keeping for QC programmes of therapy units and of U calculations.

The purpose of the documentation of U calculations can be summarized in the following practical objective: *the records shall make it possible for an experienced clinical physicist, not from the same department, to verify at some time in the future, that the prescribed dose was administered to the patient assuming that the therapy unit was in perfect working order (the subject of technical quality control) and that no setup errors were made (one of the subjects of departmental quality assurance).* Such an audit includes verification of the data used to describe the beam, the influence of any accessory that may be used, the correctness of the mathematical relations used to calculate the monitor setting (or the time setting) of each irradiation, and the absence of numerical errors. Archiving records is not just a question of saving data. The records must be intelligible, and it is the responsibility of the department to develop procedures and maintain standards so that an external check or audit can be performed in a relatively simple way.

9.4.2. Beam Data and Formulae

When the beam data are documented, not only will the tables of the final values be filed, but also descriptions of the techniques used for the measurements and for analysis of the results. This information can, for instance, be done in hard-bound laboratory notebooks and in internal reports. When computerized equipment is used, printouts of the data should be archived together with precise definitions of the quantities involved and with reference to original notes. While data may be filed electronically, it is often useful if hard copy printouts are stored. Such printouts, as for all tables of beam data, must be annotated with the date when they are taken into clinical use and signed by the responsible qualified physicist. Reference should be made on the data sheets to the records of measurement and to relevant computer files. No informal changes (corrections) to the data sheet should be permitted; if errors are discovered or updates needed, a new data sheet should be issued and the old one declared obsolete. One copy of the obsolete sheet must be marked with the date it was last valid and filed permanently for future reference.

The importance of defining the type of beam data precisely cannot be overemphasized. There is a great deal of inter-institutional variation, and it is not to be taken for granted that an independent physicist auditor will be familiar with all the details. This extends to the mathematical formulae used in the dose calculation to give the relationship between prescribed dose and monitor setting (or time). Each department must state these definitions and these formulae in a clear, complete but succinct manner for an audit to be possible. One cannot expect the auditor to do so! The resulting documents should be treated under the same document control system as described above for data sheets, e.g. provided with date when valid, signature, date when obsolete, references to other archived materials.

The requirement to ensure clear understanding of the systems used and the interpretation of any data applies equally to the manufacturers of treatment planning systems. A section in the operation manual should be devoted to these subjects, but in addition, and especially when the system permits "customizing" or other manipulation which may introduce subtle changes in definitions of concepts, formats should be available for printout of critical quantities and their relationships. The numerical values of these quantities must be easily documented and these printouts should be considered official, departmental data sheets. In addition, the trail of how they were measured and edited must be clearly marked with references.

9.4.3. Patient Calculations

When calculations are made for individual patients, sufficient information needs to be included in the permanent record to allow retrospective analysis as to which values were used, how they were determined, how they were defined exactly, and how they were used in the U calculations. If the data sheets discussed above are well organized and complete, the date of calculation will, in principle, provide a sufficient cross-reference. However, the form used for monitor unit calculation serves purposes other than documentation. It should be designed in such a manner that the calculation logic is clear. If it is intended for manual calculation, it should remind the person filling it out of factors that may be required, depending on the treatment technique. Whether computer printout or manual form, the design should facilitate double-checking, particularly of the information that is specific to the patient: depth, field size and shape, wedge, compensator, and so on. Manual calculations must be verified to eliminate numerical mistakes. Computer based calculations may not require this, provided input data are verified and the system has been thoroughly evaluated previously. The latter must, of course, also be documented.

9.4.4. Signatures and Responsibilities

The department policy should specify those persons responsible for signing the calculation form and other documentation. It is strongly recommended that a qualified, responsible physicist signs all data sheets and any documentation which relates to the mathematical basis of the calculations to reflect his or her responsibility for coordination of the many components, some of which demand a great deal of knowledge and professional judgement. This person should also have responsibility for archiving this information. The calculations for individual patients should be signed to indicate that the person generating them followed the proper current procedures and that calculations have been checked independently and shown to be correct.

APPENDIX A.1

A.1.1. RELATIONSHIP BETWEEN PERCENTAGE DEPTH DOSE AND TISSUE-PHANTOM RATIO.

Percentage depth doses have previously been referenced to the depth of dose maximum z_m , and various conversion formalisms exist which relate this physical parameters to other quantities, such as tissue-phantom ratio, tissue-air ratio, and tissue-maximum ratio (BJR, 1983). The relationship to tissue-phantom ratio using the convention that field size for P and F is defined at the source-surface distance f' , is given by :

$$P(z, s, f') = 100 \cdot \frac{T[z, s(f' + z) / f']}{T[z_m, s(f' + z) / f']} \cdot \frac{V[z_m, s(f' + z) / (f' + z_m)]}{V(z_m, s)} \cdot \left[\frac{f' + z_m}{f' + z} \right]^2 \quad (A.1.1)$$

If the reference depth is equal to the depth of dose maximum (z_m), this equation becomes :

$$P(z, s, f') = 100 \cdot T[z, s(f' + z) / f'] \cdot \frac{V[z_m, s(f' + z) / (f' + z_m)]}{V(z_m, s)} \cdot \left[\frac{f' + z_m}{f' + z} \right]^2 \quad (A.1.2)$$

In this report, percentage depth dose and tissue-phantom ratio are normalised to a reference depth z_R . Under these circumstances, the above equations become :

$$P_R(z, s, f') = 100 \cdot T(z, s') \cdot \frac{V(z_R, s')}{V(z_R, s'')} \cdot \left(\frac{f + z_R}{f + z} \right)^2 \quad (A1.3)$$

where s is the field size at the surface,

s' is the field size at the depth z_R , i.e. $s(f + z_R)/f$

s'' is the field size at the depth z , i.e. $s(f + z)/f$

Derivation of this equation may be seen by reference to figure A.1.1. The same equation may be re-ordered to calculate percentage depth doses from tissue-phantom ratios.

Recently Bjärnagård et al. (1996) have shown the errors occurring when tissue phantom ratios are calculated from percentage depth doses both being normalised at a constant depth close to the depth of maximum and concluded that the uncertainty is

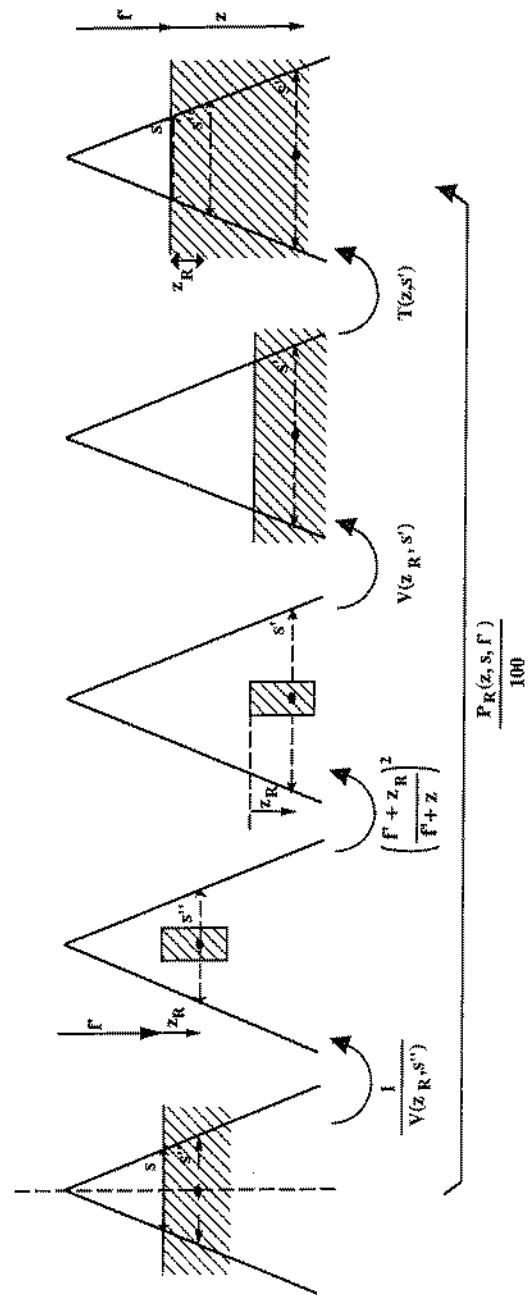


Fig. A.1.1 : Diagram illustrating the relationship between the reference percentage depth dose $P_R/100$ and the tissue-phantom ratio T .

comparable to the uncertainty in direct measurement when both quantities are normalised at a depth that provides electron equilibrium with some margin.

A.1.2. CALCULATION OF PERCENTAGE DEPTH DOSES FOR CHANGE IN SSD.

Fixed SSD treatments are normally performed at a treatment distance (f'_T) the same as that of the reference conditions under which the beam parameters were determined (f_R). When the treatment SSD differs from the reference SSD, the reference depth doses must be corrected to generate values which are applicable to the treatment SSD.

For percentage depth doses normalised to the depth of dose maximum, use may be made of published references for correcting for changes in SSD (BJR, 1983). For normalisation of depth doses at a different depth, these published references are no longer strictly applicable.

The percentage depth dose at the treatment distance f'_T may be expressed in terms of the reference percentage depth dose at the reference distance f_R using the above formula :

$$P_R(z, s, f'_T) = P_R(z, s, f'_R) \cdot \frac{T(z, s'_T)}{T(z, s'_R)} \cdot \frac{V(z_R, s'_T)}{V(z_R, s'_R)} \cdot \frac{V(z_R, s''_R)}{V(z_R, s''_T)} \cdot \left[\left(\frac{f'_T + z_R}{f'_T + z} \right) \cdot \left(\frac{f'_R + z}{f'_R + z_R} \right) \right]^2 \quad (\text{A.1.4})$$

where subscript R refers to the reference treatment distance (at which the depth dose data are available) and subscript T refers to the new treatment distance, as shown in figure A.1.2.

The above relationship is dominated by the final term for megavoltage radiation and to a first approximation the other terms may be ignored. This leads to a simplification :

$$P_R(z, s, f'_T) = P_R(z, s, f'_R) \cdot \left[\left(\frac{f'_T + z_R}{f'_T + z} \right) \cdot \left(\frac{f'_R + z}{f'_R + z_R} \right) \right]^2 \quad (\text{A.1.5})$$

Use of the simplified equation will introduce inaccuracies of less than 1 % for most megavoltage radiation fields for changes in SSD from 80 cm to 150 cm .

The above simplified equation is identical to the application of a straightforward inverse square correction in the conversion of depth doses normalised to z_m , given that :

$$P_R(z, s, f'_R) = \frac{P(z, s, f'_R)}{P(z_R, s, f'_R)} \cdot 100 \quad (\text{A.1.6})$$

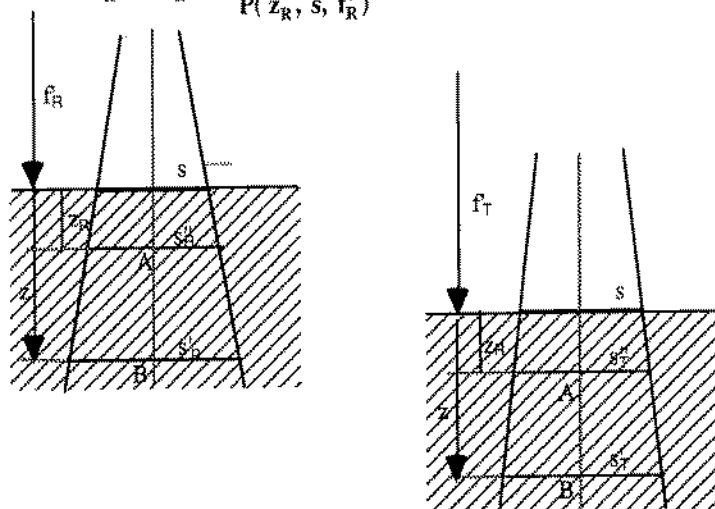


Fig. A.1.2 : Diagram illustrating the definition of the percentage depth doses at the reference source-skin distance f'_R and at the treatment source-skin distance f'_T . The reference percentage depth dose $P_R/100$ at depth z and for a field size s is defined as the ratio of the dose at point B to that at point A.

A.1.3. RELATIONSHIP BETWEEN THE REFERENCE DOSES PER MONITOR UNIT UNDER ISOCENTRIC AND FIXED SOURCE-SKIN DISTANCE CONDITIONS.

In chapters 5 and 6, the relations are given for open and modified (blocked and wedged) beams by considering two different sets of reference conditions corresponding to the two main irradiation set-ups : isocentric conditions and fixed source-skin distance.

In isocentric conditions, the reference dose per monitor unit $\dot{D}_R$ represents the dose per monitor unit measured at a point located on the beam axis at the reference depth z_R at a distance f_R from the source. The reference field size s_R , is defined at f_R (see chapter 5).

In fixed source-skin distance conditions, the reference dose per monitor unit $\dot{D}_R$ is also assumed to be measured at the reference depth z_R but the reference

conditions of distance from the source (f'_R) and field size (s_R) are taken at the surface of the phantom.

A.1.3.1 Relation when $f'_R = f_R = \text{SAD}$

In usual treatment conditions for which f'_R and f_R are taken equal to the source-isocentre distance, ($f'_R = f_R = \text{SAD}$) the reference field size s_R corresponds to the collimator opening c_R and the relation between $\dot{D}'_R$ and $\dot{D}_R$, as given in figure A.1.3, is:

$$\dot{D}'_R(z_R, c_R, f'_R) = \dot{D}_R(z_R, c_R, f_R) \cdot \left(\frac{f_R}{f_R + z_R} \right)^2 \cdot \frac{V(z_R, s''_R)}{V(z_R, c_R)} \quad (\text{A.1.7})$$

where $s''_R = c_R \cdot \frac{f_R + z_R}{f_R}$ and $\frac{V(z_R, s''_R)}{V(z_R, c_R)}$ is the ratio of the volume scatter ratios for the field sizes s''_R and c_R , respectively. It is equal to the ratio of the output ratios $\frac{O_R(s''_R)}{O_0(s''_R)}$.

A.1.3.2 Relation when $f'_R \neq f_R$

If f'_R is different from f_R and under the condition that the reference field size s_R is the same at f_R and at f'_R corresponding to the collimator openings c'_R and c''_R , respectively, the relation between $\dot{D}_R$ and $\dot{D}'_R$ (see Fig. A.1.4), becomes:

$$\dot{D}'_R(z_R, s_R, f'_R) = \dot{D}_R(z_R, s_R, f_R) \cdot O_0(s'_R) \cdot \left(\frac{f_R}{f'_R + z_R} \right)^2 \cdot \frac{V(z_R, s''_R)}{V(z_R, s_R)} \quad (\text{A.1.8})$$

with $s''_R = s_R \cdot \frac{f'_R + z_R}{f'_R}$

$$s'_R = s_R \cdot \frac{f_R}{f'_R}$$

$O_0(s'_R)$ is the output ratio in the mini-phantom for the field size s'_R .

$\frac{V(z_R, s''_R)}{V(z_R, s_R)}$ is the ratio of the volume scatter ratios for the field sizes s''_R

and s_R respectively.

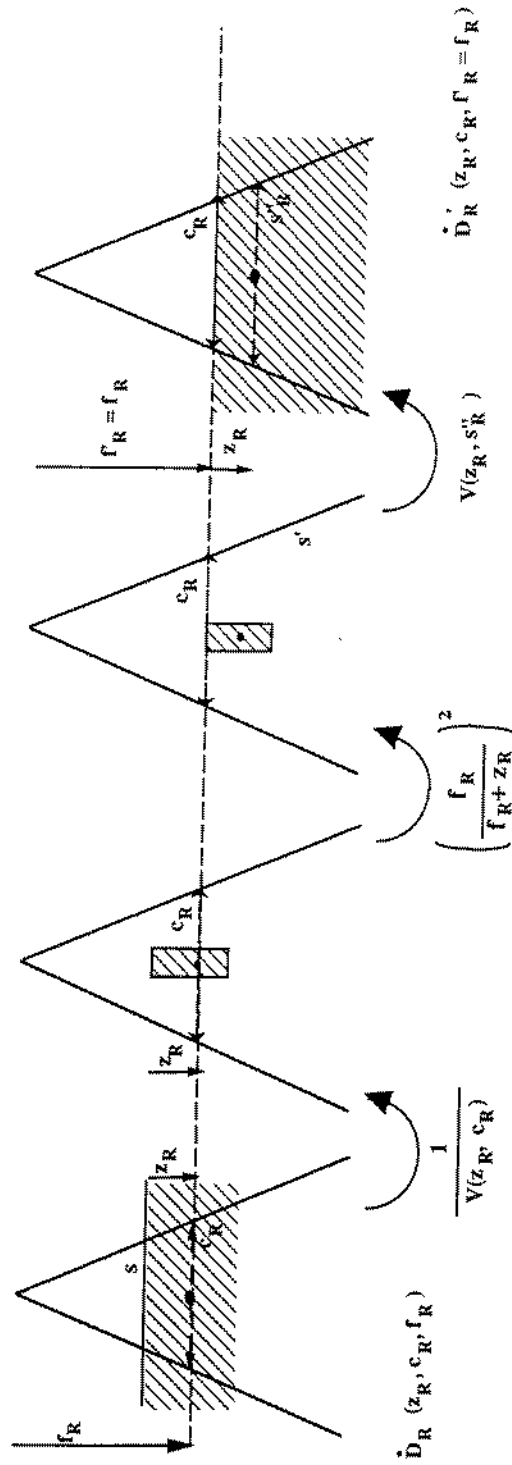


Fig.A.1.3 : Diagram illustrating the relationship between the reference doses per monitor unit under isocentric and fixed source-skin distance conditions in the case where the reference distances f_R and f'_R are identical (generally taken as equal to the source-isocentre distance). The field sizes at f_R and af'_R are assumed to be equal to c_R .

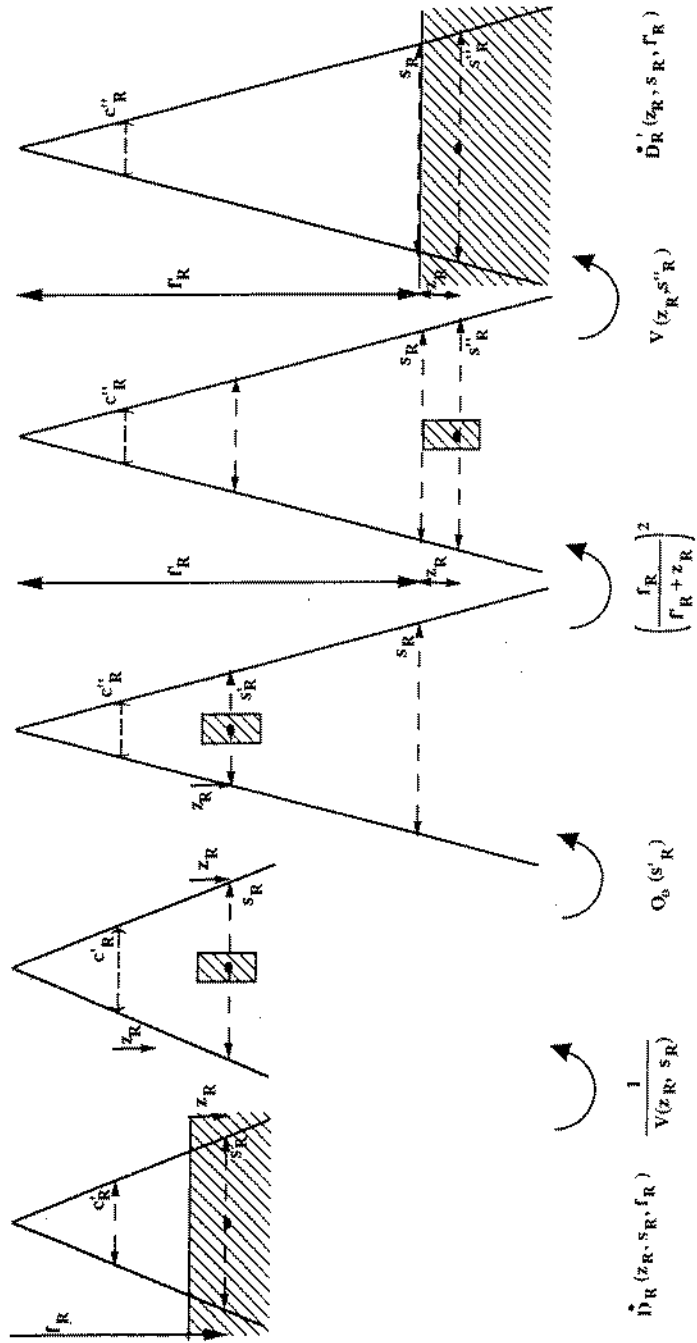


Fig.A.1.4 : Diagram illustrating the relationship between the reference doses per monitor unit under isocentric and fixed source-skin distance conditions in the case where the reference distances f_R (under isocentric conditions) and f'_R (under fixed source-skin distance conditions) are different. The field sizes at f_R and at f'_R are assumed to be equal to s_R .

APPENDIX A.2

MONITOR UNIT CALCULATION IN WEDGED BEAMS

A.2.1. INTRODUCTION

In this report, no mention is made about dynamic wedges. Only wedged beams resulting of the use of physical wedges are considered. These wedged fields are usually obtained in two ways in clinical practice :

- In the first method, a specific wedge filter is used during the whole treatment to provide the required isodose angle. The wedge filter is usually fixed at the end of the collimator in a support which ensures a reproducible position in the beam. The method implies the availability of several wedge filters of different angles in order to achieve the different isodose angles.

- The second possibility is offered in treatment units equipped with a unique wedge filter of a large angle (usually 60°) incorporated in the head of the treatment unit. Such a motorised wedge filter is moved in the beam during part of the irradiation. An appropriate choice of the time with the wedge filter in the beam allows to obtain the required isodose angle θ . θ -values up to the maximum wedge angle can be achieved in practice.

The calculation of the number of monitor units to be delivered during a treatment will be considered separately for each of the two possibilities.

A.2.2. USE OF A FIXED WEDGE DURING THE WHOLE TREATMENT SESSION

The number of monitor units needed to deliver the prescribed dose D_P at the point of interest in the treatment conditions can be calculated directly from one of the different relations proposed in section 6.2 for various treatment techniques (isocentric, fixed SSD).

A.2.3. USE OF A WEDGE DURING PART OF THE TREATMENT SESSION.

A.2.3.1 Monitor Unit Calculation

In practice, each treatment session combines U_w monitor units given with the wedge in the beam and U_n monitor units given without the wedge.

The ratio of U_w to U_n is chosen in order to fulfil the two following conditions:

- obtaining the prescribed isodose angle,
- delivering the prescribed total dose D_p as the sum of the doses with, D_w , and without, D_n , the wedge in the beam.

The determination of the number monitor units U_w and U_n is based on the following relations :

$$\begin{cases} D_p = D_w + D_n \\ \frac{D_w}{D_p} = B \end{cases} \quad \text{with} \quad \begin{cases} D_n = U_n \cdot \dot{D}_n \\ D_w = U_w \cdot \dot{D}_w \end{cases} \quad (\text{A.2.1})$$

where $\dot{D}_n$ and $\dot{D}_w$ represent the absorbed dose per monitor unit in treatment conditions without and with the wedge and B is the fraction of the prescribed dose which is delivered with the wedge filter in the beam.

From the relations A.2.1, the number of monitor units to be delivered in the wedged and open fields can be expressed by

$$\begin{cases} U_w = \frac{B}{\dot{D}_w} \cdot D_p \\ U_n = \frac{1-B}{\dot{D}_n} \cdot D_p \end{cases} \quad (\text{A.2.2})$$

Lastly, the ratio of the number of monitor units delivered with the wedge U_w to the total number of monitor units U_T can be given by the relation :

$$\frac{U_w}{U_T} = \frac{U_w}{U_w + U_n} = \frac{B}{B + (1-B) \cdot \frac{\dot{D}_w}{\dot{D}_n}} \quad (\text{A.2.3})$$

The expressions of $\dot{D}_n$ and $\dot{D}_w$ are given in chapters 5 and 6.

A.2.3.2. Expression of the Fraction of the Prescribed Dose Delivered with the Wedge

Different mathematical models have been proposed in the literature for combining open and wedged fields. A simple approximate model was proposed by Thatcher (1970) who suggested that the effective wedge angle θ_E resulting from the addition of the two beams (with and without wedge) is equal to the nominal wedge angle θ_w for the wedged beam weighted by the fraction of the wedged field B .

$$\theta_E = \theta_w \cdot B \quad (\text{A.2.4})$$

This approximation can be considered as reasonable for θ_w values less than 45° but is inadequate in situations where the nominal wedge angle θ_w is 60° .

A more general and accurate theoretical model was proposed by van der Laarse et al. (1984) and later by Petti and Siddon (1985). These authors have introduced the ratio of $\tan\theta_E / \tan\theta_w$ and have taken into consideration the differences in the central axis depth dose curves between the wedged and open fields caused by the beam hardening introduced by the wedge.

The general expression for B given by Petti and Siddon is:

$$B = \frac{e_w}{\frac{\tan\theta_w}{\tan\theta_E} + e_w - 1} \quad (\text{A.2.5})$$

where e_w can be equated to the ratio of the effective attenuation coefficients of these two fields (open and wedged).

The parameters θ_w and e_w depend on depth, energy and field size and their values must be determined experimentally. Since, for the field sizes smaller than 20 cm x 20 cm, as generally encountered in clinical practice, the parameter e_w has been found very close to unity, the above expression of B can be approximated by:

$$B = \frac{\tan\theta_E}{\tan\theta_w} \quad (\text{A2-6})$$

The predictions based on this model agree with experimental data for the Philips SL75 6 MV accelerator to within 3% in the range of field sizes considered (Petti and Siddon, 1985).

A.2.3.3 Application to a Treatment Performed under Isocentric Conditions.

Let us consider a treatment performed under isocentric conditions with motorised wedge, having a wedge angle θ_w (usually about 60°).

A dose D_P at the point of interest is required, with a dose distribution characterised by an angle θ_E . This distribution is produced by a combination of open and wedged beams of the same field size. The point of interest is considered to be at a depth z and at a distance f_R from the source equal to the source-rotation axis distance of the treatment unit. At f_R , the side of the treatment equivalent field size is c .

According to equation A2-3, the fraction U_w of the total number of monitor units

($U_n + U_w = U_T$) to be delivered with the wedge in the beam is :

$$\frac{U_w}{U_T} = \frac{U_w}{U_w + U_n} = \frac{B}{B + k_w(z, c) \cdot (1 - B)} \quad (A.2.7)$$

where $k_w(z, c)$ represents the wedge factor at depth z , and for the field size c under consideration.

Considering, as a first approximation, that $B = \tan \theta_E / \tan \theta_w$ one can write :

$$\frac{U_w}{U_T} = \frac{\tan \theta_E}{k_w(z, c) \cdot \tan \theta_w + [1 - k_w(z, c)] \cdot \tan \theta_E} \quad (A.2.8)$$

This expression of U_w / U_T is similar to that reported by van der Laarse et al. (1984). It can be applied in practice for a given wedge isodose angle $\theta_E < \theta_w$.

Example :

A treatment is performed on a treatment unit having a motorised wedge with a wedge angle $\theta_w = 60^\circ$. The wedge factor $k_w(z, c)$ at the point of interest for the field size under consideration is 0.25. The required wedge angle is $\theta_E = 30^\circ$.

In order to reach this objective, each treatment session must be performed by combining the open field and the wedged field in such a way that the ratio of the number of monitor units with the wedge filter U_w to the total number of monitor units U_T is :

$$\frac{U_w}{U_T} = \frac{\tan 30^\circ}{0.25 \cdot \tan 60^\circ + (1 - 0.25) \cdot \tan 30^\circ} = 0.667 \quad (A.2.9)$$

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