A simple Formalism for Calculation and Verification of Dose in Asymmetric X-ray Fields

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Abstract

A simple method for dose calculations in the center of asymmetrically collimated fields is presented. We use existing base data for symmetric fields and a measured correction factor that allows for off-axis variations of the incident primary beam fluence and beam quality. Our model does not require any considerations of collimator and phantom scatter factors. This algorithm is verified by comparing its predictions with measurements of a series of asymmetric fields at a Clinac 600C with 6 MV photons and a Clinac 2300C/D with 6 and 18 MV photons. Calculated and measured doses agree mostly within 1% for all energies. Mean deviations were found to be less than 0.5%; a maximum deviation of 1.09% was observed in one case.

1 Introduction

Modern linear accelerators are equipped with collimators that allow independent setting of each jaw. The application of asymmetrically collimated beams has become increasingly common in many clinical standard situations [1]. In the case of asymmetric jaw positions, the center of the field is off the central axis of the collimator. However, the basic dosimetric data required for dose calculations (PDD, TMR, scatter factors, etc.) are normally acquired on the central axis for a limited number of square and rectangular symmetric fields. For independent jaws the number of possible combinations of jaw positions is considerably increased. Consequently the basic dosimetric data cannot be measured directly by a reasonable number of measurements. Although various methods has been discussed in the literature [2–15], there is no generally accepted procedure to determine the absorbed dose in the center of an asymmetric field. In this paper we propose a dose calculation method which employs symmetric beam data
and a correction factor that is a function of the off-axis distance (OAD) of the field center and the depth of the point of interest.

2 Methods and Materials

2.1 Theoretical

It has been pointed out in earlier works that the difference in dose between a symmetric and an asymmetric field of the same size is mainly caused by the existence of the beam flattening filter which leads to variations in primary photon fluence and energy spectrum with the off-axis distance [2,3]. This implies that the change of dose output in an asymmetric field can be described by an off-axis correction factor. To consider variations of the primary photon fluence, the correction factor may be obtained by using the primary off-axis ratio (POAR) measured in air [4]. However, this must be modified to include the effect of off-axis beam quality variations, resulting in radiation that is more penetrating on-axis than off-axis. Many authors proposed to use the off axis ratio (OAR) derived from beam profiles measured in a water phantom at the depths of interest for the largest field size available [2,10]. The OAR is, however, strongly influenced by phantom scatter concealing the primary (i.e. unscattered) component of the dose (see Fig. 1). For that reason separation of primary dose is essential to isolate the effects of off-axis beam changes. The primary dose component is usually obtained by dose measurements in narrow beam geometry and extrapolation to zero field size. However, large uncertainties arise from the process of nonlinear extrapolation because practical measurements in high-energy photon beams are restricted to field sizes larger than approximately $3 \times 3 \text{ cm}^2$ by requirement of lateral electron equilibrium [18]. At these field dimensions, the total scatter factor is changing drastically and the extrapolation to zero field size is very uncertain. In this work we follow a different approach that allows extraction of primary dose without any extrapolation.

The central axis dose $D(X \times Y, 0, 0, d)$ to a point $P(0, 0, d)$ in a symmetric field of fieldsize $X \times Y$ can be written as

$$D(X \times Y, 0, 0, d) = D_{ref} \cdot S_{c,p}(X \times Y) \cdot TMR(X \times Y, d) \quad (1)$$

where $D_{ref}$ is the reference dose measured for a reference field size at the depth of the dose maximum $d_{max}$, $TMR(X \times Y, d)$ is the tissue maximum ratio at depth $d$ for field size $X \times Y$ and $S_{c,p}(X \times Y)$ is the total scatter factor (output factor) at $d_{max}$ of field size $X \times Y$ normalized to 1.0 for the reference field size. In practice, the reference field size is usually chosen to $10 \times 10 \text{ cm}^2$ but Eq.
(1) is valid for arbitrary reference field sizes, even for the zero-area field size limit. In this case, the reference dose $D_{\text{ref}}$ equals the primary dose $D_{\text{pr}}$ due to photons that have not been scattered in either the collimator or the phantom. Previous measurements indicate that, within experimental uncertainties, the scatter factors are independent of the off-axis position[2,13]. Therefore, if we consider ratios of doses at the center of identical fields positioned at an off-axis point $(x, y)$ and on the central collimator axis, the scatter factor is cancelled:

$$\frac{D(X \times Y, x, y, d)}{D(X \times Y, 0, 0, d)} = \frac{D_{\text{pr}}(x, y, d_{\text{max}})}{D_{\text{pr}}(0, 0, d_{\text{max}})} \cdot \frac{TMR(X \times Y, x, y, d)}{TMR(X \times Y, 0, 0, d)}$$

$$= \frac{D_{\text{pr}}(x, y, d_{\text{max}})}{D_{\text{pr}}(0, 0, d_{\text{max}})} \cdot C(x, y, d) = OAR_{\text{pr}}(x, y, d)$$  \hspace{1cm} (2)
$TMR(X \times Y, x, y, d)$ and $TMR(X \times Y, 0, 0, d)$ denote the off-axis and central axis TMR, respectively. Although TMR is related to the field size, we assume that the ratio $C(x, y, d) = \frac{TMR(X \times Y, x, y, d)}{TMR(X \times Y, 0, 0, d)}$ does not vary significantly with the field dimensions $X \times Y$ and thus represents the change in primary beam attenuation with the off-axis distance. The underlying assumption in this approximation is that off-axis changes in beam quality affect the primary beam transmission and the scattered dose in the same way. This is in contrast to the approaches of Kepka, Loshek, Khan and Gibbons et al. [16,17,2,8] who assumed that the scattered component of dose is independent of the beam energy.

2.2 Experimental

The measurements were performed with a Varian Clinac 600C (6 MV photons) and a Clinac 2300 C/D (6 and 18 MV photons). Both treatment units are equipped with independent jaws assigned to as Y1 and Y2 for the upper and X1 and X2 for the lower jaws. The travel ranges are +20 to -10 cm for the Y jaws and +20 to -2 cm for the X jaws, respectively. Therefore the coordinates $(x, y)$ and the off-axis distance (OAD) of the beam center can be written as

$$x = \frac{1}{2}(X2 - X1)$$

$$y = \frac{1}{2}(Y2 - Y1)$$

$$OAD = \sqrt{x^2 + y^2}$$

A WP700 water tank system (Wellhoefer Dosimetrie, Germany) and a 0.3 cm$^3$ cylindrical ionization chamber (PTW Type 31003) connected to a PTW Unidos dosemeter (Physikalisch-Technische Werkstaetten, Germany) were used for the measurements. A $5 \times 5$ cm$^2$ square field was moved through the water tank by fixing the X jaws to $X1 = X2 = 2.5$ cm and successively changing the position of the Y jaws from $Y1 = Y2 = 2.5$ cm, i.e. $(x, y) = (0, 0)$, to $Y1 = -9.5$ cm and $Y2 = 14.5$ cm, i.e. $(x, y) = (0, 12$ cm), in increments of 1 cm. Due to the overtravel limit of -10 cm for the Y jaws, a maximum OAD of y=12.5 cm can be achieved with a field size of $5 \times 5$ cm$^2$. For larger distances, the field size was increased successively by fixing the Y1 jaw at -10 cm and changing the Y2 jaw position to 16, 18 and 20 cm resulting in OAD of y=13, y=14 and y=15 cm, respectively. The X jaws were simultaneously increased to retain a square field. The ionization chamber readings at the beam centers were recorded at depths of 5, 10, 15 and 20 cm with fixed source-detector distance of 100 cm. $OAR_{pr}(0, y, d)$ at an off-axis point $(0, y)$ was calculated as
the ratio of chamber readings measured in the field centered at \((0, y)\) and at the central axis \((0, 0)\). The procedure is illustrated in Fig. 2.

\(OAR_{pr}(0, y, d)\) applies strictly to an off-axis point on the \(y\)-axis, i.e. for asym-

![Fig. 2. Schematic view of the experimental setup. The dose is measured at the center of a small scan field that moves along the \(y\)-axis through the water phantom. \(OAR_{pr}\) is normalized to the doserate obtained when the field center is positioned on the collimator axis. The source-detector distance is fixed to 100 cm.](image)

metric setting of the \(Y\) jaws only. However, imperfections in beam steering and flattening may result in a different \(OAR_{pr}\) along the \(x\)-axis. Due to the travel limits of the \(X\) jaws, \(OAR_{pr}(x, 0, d)\) cannot be measured in the same way as described above. Therefore \(OAR_{pr}(x, 0, d)\) is derived from \(OAR_{pr}(0, y, d)\) by means of normalization:

\[
OAR_{pr}(x, 0, d) = \frac{OAR(x, 0, d)}{OAR(0, y, d)} \cdot OAR_{pr}(0, y, d)
\]  \(3\)

where \(OAR(x, 0, d)\) and \(OAR(0, y, d)\) are obtained from beam profiles along the \(x\)- and \(y\)-axis, respectively, measured in water for the maximum opening of the collimator, i.e. a \(40 \times 40\) cm\(^2\) square field. These profile scans were made in a WP700 water tank at depths of 5, 10, 15 and 20 cm using two 0.14 cm\(^3\) cylindrical ionization chambers (Type IC10, Wellhoefer) connected to a WP5006 dual channel electrometer. One channel was used for the reference
signal and the other one for the field signal. The source-detector distance was fixed to 100 cm for each scan.

For asymmetric setting of both the X- and the Y-jaws, the center of the resulting field is shifted to a point \((x, y)\) which is off both of the principal collimator axes. In this more general case, the corresponding correction factor \(OAR_{pr}(x, y, d)\) is derived by weighted averaging of \(OAR_{pr}(x, 0, d)\) and \(OAR_{pr}(0, y, d)\):

\[
OAR_{pr}(x, y, d) = \frac{1}{OAD^2} \cdot (x^2 \cdot OAR_{pr}(x, 0, d) + y^2 \cdot OAR_{pr}(0, y, d))
\]

The absorbed dose in the center \((x, y)\) of an asymmetric field can therefore be expressed as

\[
D(X \times Y, x, y, d) = OAR_{pr}(x, y, d) \cdot D(X \times Y, 0, 0, d)
\]

where \(X = X_1 + X_2\) and \(Y = Y_1 + Y_2\) are the width and length of the field and \(D(X \times Y, 0, 0, d)\) is the central axis dose for the symmetric field given by Eq. (1).

3 Results

Figures 3 (a) - (f) show measured \(OAR_{pr}\) along the x- and y-axis for the Clinac 2300C/D (6 and 18 MV photons) and the Clinac 600C (6 MV photons) as a function of depth. As expected, \(OAR_{pr}\) first increases with increasing off-axis distance which is according to the fluence distribution of the incident beam. Due to increasing beam attenuation at larger distances, the gradient decreases and \(OAR_{pr}\) reaches some kind of plateau or even falls after passing through a maximum. This effect is more significant at larger depths, where absorption is more dominant. For the Clinac 600C there is no significant difference in the distribution of \(OAR_{pr}(0, y, d)\) and \(OAR_{pr}(x, 0, d)\). In case of the Clinac 2300C/D, however, \(OAR_{pr}(0, y, d)\) is typically higher than \(OAR_{pr}(x, 0, d)\) by 0.5% to 1%. This is a consequence of the different beam steering systems of the two Clinac types. The Clinac 2300C/D has a beam bending system, which may lead to different primary fluence profiles in and cross the bending plane.

In order to check the reliability of the proposed dose calculation method, Eq. (5) was applied to a number of clinically relevant asymmetric fields and compared to measurements. The measurements were made by using the same experimental equipment as for the determination of \(OAR_{pr}(0, y, d)\). For each
Fig. 3. Measured $OAR_{pr}$ along the y-axis and the x-axis for (a) and (b): Clinac 2300C/D with 6 MV x-rays, (c) and (d): Clinac 2300C/D with 18 MV x-rays, (e) and (f): Clinac 600C with 6 MV x-rays.

jaw configuration, the dose was measured at depths of 5, 10 and 20 cm. The chamber readings in the center of the asymmetric fields were normalized to the readings obtained for the corresponding symmetric fields of the same dimensions. Table 1 gives a comparison of the measured ratios of chamber readings with those predicted by Eq. (5).

Although $OAR_{pr}$ was derived from isocentric measurements, it should be noted that Eq. (5) is also applicable for fixed SSD treatments, if the off-axis coordinates $(x, y)$ are chosen at the entry level of the beam. Experimental verification for asymmetric setting of the Y jaws at the Clinac 2300C/D is given in Table 2. For these measurements the source-surface distance was fixed to 100 cm and the chamber was positioned at the center of the beam projection at the depth of interest.

Calculated and measured dose ratios are in good agreement with an error averaged over all beam qualities, jaw settings and depths of less than 0.5% and a maximum error for all cases of 1.09%.
Table 1
Comparison of measured and calculated ratios of chamber readings between asymmetric and symmetric fields at the beam center. The source-detector distance was fixed to 100 cm (isocentric treatment).

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<th>X2 (cm)</th>
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Mean 0.25 0.45 0.34
Max. 0.66 0.96 1.09

4 Discussion

Asymmetric collimation does lead to significant errors (up to approximately 7%) in dose calculations if changes in primary beam intensity and beam quality with the off-axis distance are ignored. A model has been presented which explains the radial variation of primary dose as a result from differential beam

8
Table 2
Comparison of measured and calculated ratios of chamber readings between asymmetric and symmetric fields at the beam center. The source-surface distance was fixed to 100 cm (SSD treatment).

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Mean 0.35  0.35
Max. 0.85  0.76

hardening in the flattening filter. It has been pointed out that calculation of dose to a point at the center of an asymmetric field involves the use of the off-axis ratios of the primary dose component (\(OAR_{pr}\)) as a function of the depth. Since off-axis beam quality variations are incorporated in \(OAR_{pr}\) no additional corrections, such as off-axis attenuation coefficients, are needed. An experimental method has been described in which \(OAR_{pr}\) can be easily extracted by using the independent collimator system of the accelerator. The good agreement of calculated and measured dose output for asymmetric fields justifies the assumptions yielding Eq. (2). The proposed formalism can be used for dose determination for asymmetric fields without invoking CT-based treatment planning systems and allows spot checks of computerized monitor unit calculations.
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