An Estimated Correction Factor for 10 MV Photon Output Due to Backscattered Radiation from Collimator Jaws Entering Into the Monitor Chamber

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ABSTRACT: An accurate dose per monitor unit is an important quantity for the determination of the amount of beam delivery and, especially, for any model-based algorithm that is used for evaluating dose calculations. The monitor unit is directly proportional to the dose deposited in the monitor chamber, which mainly comes from a forward radiation. A minor dose is due to a backward radiation from the collimator jaws and the level of this depends on the position of the jaws and the field size. In this study, the photon outputs were measured for square field sizes ranging from 5 x 5 up to 30 x 30 cm^2. They were also determined by the Monte Carlo method in which the linac model was first validated in details by comparison of the percentage depth doses and the dose profiles. Then, the doses at the reference depth for various field sizes were extracted and used to calculate the photon outputs. The estimated amount of backscattered radiation from the collimator jaws into the monitor chamber was performed by comparison between the photon outputs from both methods. Since the monitor unit was not a concern in the Monte Carlo simulation, the difference between the measured and the simulated photon output would principally be due to this backscattered radiation, and ranged from 1 to 3%. As a result, correction factors for the calculated photon output for each studied field size can be evaluated. We found that they were in excellent agreement with the numerical values obtained by using the formula reported by Liu *et al*.[10] Estimation of and correction for the backscattered radiation from the collimator jaws to the monitor chamber is necessary for an accurate dose determination using the Monte Carlo method.

Key words: Backscattered radiation, photon output, Monte Carlo

INTRODUCTION

Generally, the amount of therapeutic beam delivered from a medical linear accelerator is determined by the dose or energy deposited in a monitor chamber. Typically, the dose in the monitor chamber principally comes from the traversing beam but a small part of it results from the beam scattering from the collimator jaws that defines the field shape. This backscattering must be taken into account in order to achieve an accurate dose calculation, especially for clinical linear accelerators in which that the distal monitor chamber window is thin[1] and, also in planning a modern treatment that uses a convolution algorithm or other model-based methods.[2,3] The accuracy of the dose delivered to the patient depends on the precision of the monitor unit's computation. The use of advance beam delivery techniques, which involve several complex fields, such as IMRT, using dynamic jaw movements requires a high accuracy of determination in the monitor unit for each component field.

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The direct measurement of the backscatter radiation into the monitor chamber is not a practical approach because of the difficulty in distinguishing the small backscattered dose from the major dose of the forward beam.[4-9] Moreover, measuring the backscattering dose for every field component takes additional time and manpower. An effective solution is perhaps the use of a Monte Carlo simulation technique.[10,11] The ability to identify the dose contribution from each linac's component can be done in the Monte Carlo simulation. Liu *et al.*[10] simulated the 10 MV photon beams delivered from a Varian Clinac 2100C and determined the backscatter radiation levels for various positions of the collimator's jaw. They derived a formula to determine the amount of relative backscatter radiation from the collimator jaws into the monitor chamber as a function of the jaws' position. Their formula is very useful since it is applicable for any jaw opening, including asymmetric fields and fields using dynamic jaw movement.

In this study, we investigated the 10 MV photon outputs from the measurement and from the Monte Carlo simulation. Direct comparison of the two datasets were used to estimate a correction factor for to the backscatter radiation from the jaws into the monitor chamber. In the Monte Carlo simulation, the beam model was verified in details for various square field sizes. Common dose distributions in a water phantom, such as the percent depth doses and dose profiles, for the studied field sizes were matched between the calculated and the measured results. After they were found to be in good agreement, the photon outputs were estimated in the simulation. Any discrepancy between the photon outputs obtained by both methods was identified as a result of backscatter radiation from the jaws into the monitor chamber, since it affected the monitor unit in the measurement only. In the simulation, no monitor unit was defined. Thus, a correction factor for the simulated photon output was extracted. Our correction factors were compared with those calculated by using the backscatter radiation from the formula in Liu *et al.*[10]

METHODS AND MATERIALS

To obtain the photon outputs from the measurement, the doses at 10 cm depth from a 10 MV photon beam with one of six different field sizes (5 x 5, 7 x 7, 10 x 10, 15 x 15, 20 x 20 and 30 x 30 cm^2) were measured in a water tank at 100-cm SSD. The photon beams were generated from a Clinac 2100C medical linear accelerator (Varian Oncology Systems, Palo Alto, CA). The point dose measurement for each field size was done using a farmer type ionization chamber NE2571 (Nuclear Enterprise Limited). Its active volume is 0.69 cm^3

with a volume's length of 24.1 mm and a wall thickness of 0.36 mm. The statistical uncertainty of the ionization chamber measurement was estimated to be 0.5% or less. The output factors were calculated relative to the dose with a 10 x 10 cm^2 field.

To estimate the output factors from the Monte Carlo simulation, the model of the accelerator must be constructed and validated. The machine simulation was done using the EGSnrc user code, BEAMnrc.[12] The linac head was modeled according to the detailed specifications supplied by the manufacturer. The model included an x-ray target, a primary collimator, a flattening filter, a monitor chamber, a field light mirror and an upper and lower jaw (X and Y jaws). The unknown parameters in the simulation were the characteristics of the electron beam incident on the x-ray target, that is the mean energy, energy distribution and intensity distribution. Here, the pencil electron beam was assumed to be monoenergetic, monodirectional and circularly parallel with the central axis. In addition, the energy distribution was not taken into account since it has negligible influence on the beam characteristics.[13,14] The intensity distribution was assumed to be Gaussian, which was characterized by a full width at half maximum (FWHM). Thus, in our beam model the energy and the FWHM were the parameters of the incident electron beam. Having these parameters, the phase spaces were generated and used as a radiation source to calculate the 3D dose distribution in a water phantom using the EGSnrc user code, DOSXYZnrc. The ECUT and PCUT were set to 0.7 and 0.01 MeV, respectively. For the numerative simulation efficiency, the voxel size was varied, photon splitting was used instead of recycling, and the air between the secondary collimator and the phantom was included in the DOSXYZnrc simulation.[15] The number of histories and the photon splitting number were chosen so that the statistical uncertainty was less than 0.5% in the voxels of interest.

The incident beam's parameters were estimated from the best matches between the calculated and measured data. In this study, the matching data consisted of the central axis depth dose and the dose profiles at the depths of 10, 15 and 20 cm of a 30 x 30 cm^2 field. The central axis depth doses were compared at depths ranging from 5 to 25 cm. The calculated profiles were folded at the beam axis to become symmetrical and were also normalized at the beam axis. The profiles were compared within an off-axis distance of ± 14 cm for all three depths. The measurements of depth doses and profiles were done using the RFA-300 dosimetry system (Scanditronix-Wellhoefer GmbH, Germany). According to the manufacturer, the position accuracy of the scanning system is ± 0.5 mm.

The optimal parameters for the incident electron beam obtained from the 30 x 30 cm^2 field were verified by investigation of the depth doses and dose profiles of four other smaller field sizes; 5 x 5, 10 x 10, 15 x 15 and 20 x 20 cm^2. Once they were found to be in good agreement, the doses at 10 cm depth for all interested field sizes can be extracted from the simulation. They were acquired in a 1 x 1 x 0.6 cm^3 voxel, which is comparable to the effective volume of the ionization chamber used in the measurement of the output factors. Consequently, the photon outputs for all field sizes were calculated and compared with the measured values. The differences between these output factors were assumed to mainly come from the backscatter radiation from the collimator jaws into the monitor chamber, allowing the correction factors for the simulated photon outputs to be determined. Finally, we compared our correction factors for each field size to the computed values based on the backscatter given by the formula in Liu *et al*.[10]

RESULTS
The optimal parameters of incident electron beam

Using DOSXYZnrc, the calculated doses obtained from the phase spaces corresponding to various electron beam parameters were compared with the measured data. In our study, we chose the chi-square function as the cost function in the comparison, since it was considered to be more suitable than the mean absolute error, the slope of the difference, and the absolute difference of the penumbra edge point.[16] According to the previous studies,[13,14] the depth dose curve past the depth of maxima was mostly affected by the energy of the incident electron beam, while the lateral dose profile was sensitive to both the energy and the FWHM of the beam. Thus, we first varied the beam's energy,

while the FWHM was kept unchanged, and the central axis depth doses beyond the depth of 5 cm were inspected. The energy that minimized the cost function was then selected and used in the next step where the FWHM was varied. The comparison was done based on the cost function of the dose profiles at three depths. As a result, the optimal FWHM was obtained in this step. Lastly, the beam's energy was finely adjusted using the cost function of both the dose profiles and the central axis depth dose. We scanned the energy from 10.0 to 10.9 MeV and the FWHM from 1.0 to 1.4 mm. Our best estimate for the electron beam parameters are 10.4 MeV for the energy and 1.3 mm for the FWHM. The calculated dose data, depth dose and dose profiles, are summarized, along with the measured data, in Figure 1, where the good agreement between the two can be observed.

Dose distributions for various field sizes

The phase space files for different field size settings were obtained using the same optimal parameters. The four investigated field sizes were 5 x 5, 10 x 10, 15 x 15 and 20 x 20 cm^2. The central axis percent depth dose curves and the profiles at 10 cm depth for each field size were computed and compared with the measured data (Figures 2 and 3). Again, the agreement between the calculated and measured data is excellent. The differences in (PDDs) are within ± 0.5% for depths deeper than 5 cm. The profiles for the two larger field sizes (15 x 15 and 20 x 20 cm^2) are in good agreement with the measured profiles, even in the high dose gradient areas. For the two smaller field sizes (5 x 5 and 10 x 10 cm^2) the discrepancies are noticeable in the penumbra regions. However, they are within an acceptable range of ± 2 mm. In the low dose gradient region, the agreement is good for all tested field sizes.

Figure 1. **The (a) central axis percent depth doses and (b – d) dose profiles at a depth of (b) 10, (c) 15 and (d) 20 cm for the 30 x 30 cm² field size. The simulated results are shown as solid dots and the measured results are drawn in line.**
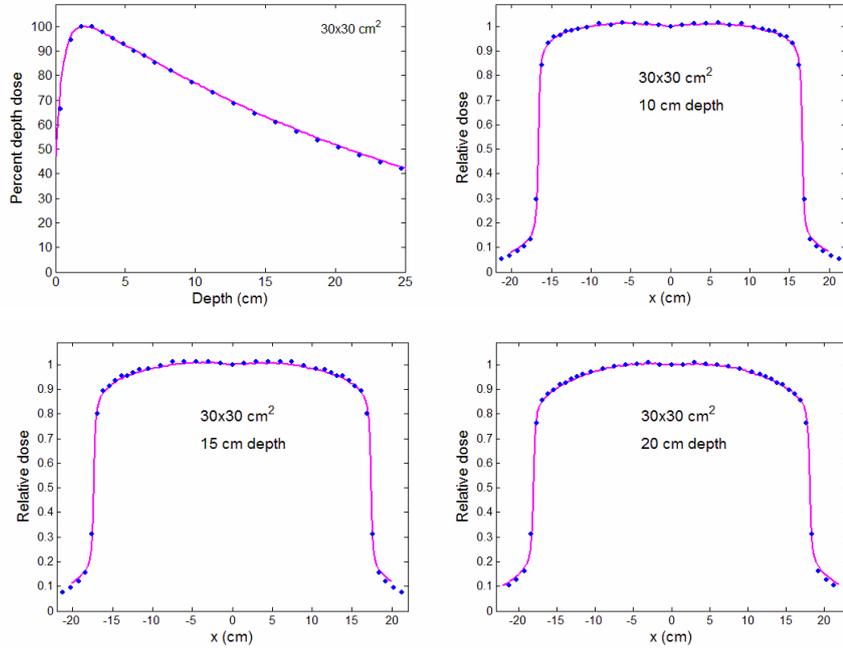

Figure 2. **The (a, c) percent depth doses and (b, d) dose profiles for the field size of (a, b) 5 x 5 and (c, d) 10 x 10 cm².**
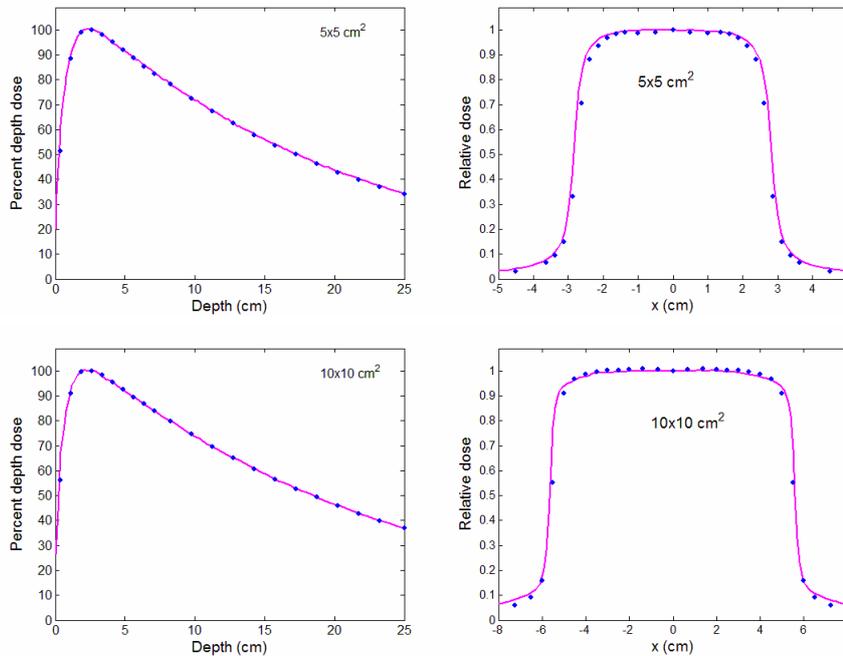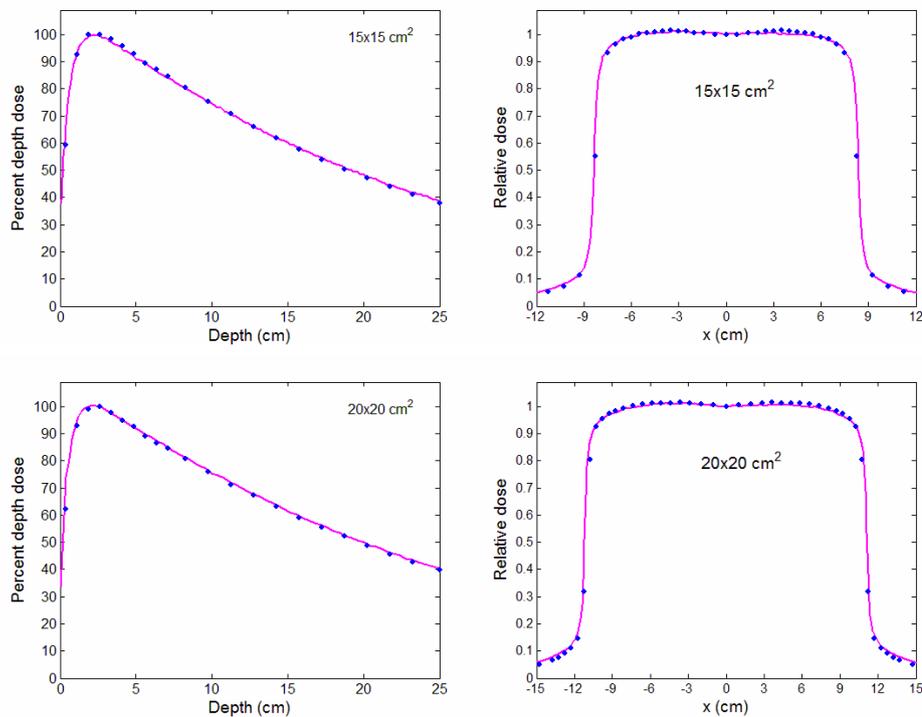
Figure 3. The (a, c) percent depth doses and (b, d) dose profiles for the field size of (a, b) 15 x 15 and (c, d) 20 x 20 cm².

The photon outputs

The doses at 10 cm depth on the central beam axis, which were obtained from calculation as well as from experimental measurement and are shown in table 1 as photon outputs, were extracted for the six field sizes of 5 x 5, 7 x 7, 10 x 10, 15 x 15, 20 x 20 and 30 x 30 cm². The dose distributions for five of these fields had been previously validated, the exception being that for the 7 x 7 cm² field size. The ratios of the measured to the calculated values are very close to the ideal of unity (Table 1). With the observed discrepancies being assumed to have originated from the backscatter from the collimator jaws into the monitor chamber.

Table 1. The 10 MV photon outputs evaluated at a 10 cm depth for the various square field sizes obtained from the measurement and calculation.

Field size	Photon output		
(cm²)	Measured	Calculated	Measured / Calculated
5 x 5	0.9145	0.9185	0.9956
7 x 7	0.9564	0.9595	0.9968
10 x 10	1.0000	1.0000	1.0000
15 x 15	1.0471	1.0445	1.0025
20 x 20	1.0762	1.0715	1.0044
30 x 30	1.1157	1.1018	1.0126

The backscatter correction S_{cb}

The backscatter correction for photon output is determined by:

S_{cb} (Ref) = [$D_f(x,y)$ / ($D_f(x,y)$ + $D_b(x,y)$)] / [$D_f(x_0,y_0)$ / ($D_f(x_0,y_0)$ + $D_b(x_0,y_0)$)].

where, D_f and D_b represent the dose from the forward and backscattered radiation, respectively. The jaw setting is denoted by (x,y), while (x_0,y_0) is the jaw setting for the reference 10 x 10 cm^2 symmetric field. However, it is more convenient to write "Scb (Ref)" in terms of the relative backscattered dose in the monitor chamber "R" because it can be obtained by the formula:

S_{cb} (Ref) = [1 + $R(x_0,y_0)$] / [1 + $R(x,y)$].

According to the formula by Liu *et al.*,[10] the amount of relative backscattered radiation $R(x,y)$ from the collimator jaws for each investigated field size is listed in Table 2. They were used to compute the backscatter correction for the photon output for each field size, labeled as S_{cb} (Ref).

Table 2. Comparison of the backscatter correction for the photon output obtained in this study (S_{cb} (Study)) and those obtained in the referred publication S_{cb} (Ref) including the amount of backscatter radiation from collimator jaws into monitor chamber computed from the formula of Liu *et al.*[10]

Field size (cm^2)	Backscattered radiation (%)	S_{cb} (Ref)	S_{cb} (Study)	Difference (%)
5 x 5	2.7976	0.9969	0.9956	-0.126
7 x 7	2.6753	0.9981	0.9968	-0.133
10 x 10	2.4829	1.0000	1.0000	0
15 x 15	2.1425	1.0033	1.0025	-0.081
20 x 20	1.7857	1.0068	1.0044	-0.240
30 x 30	1.0727	1.0140	1.0126	-0.137

According to Table 1, the ratios of the photon outputs from the measurement to the calculation are not exactly unity, except for the 10 x 10 field. Smaller fields than that have a lower ration, whilst larger fields have a ratio above unity. Basically, those ratios can be defined as the correction factor to the calculated photon outputs. Since the differences are assumed to be due the backscatter radiation, the ratios are simply the backscatter correction in our study, denoted by "S_{cb} (Study)". The comparisons between "S_{cb} (Ref)" and "S_{cb} (Study)" for each field size are also shown in Table 2, in which the difference in percentage was calculated as "[S_{cb} (Study) − S_{cb} (Ref)] *100 / S_{cb} (Ref)". The deviations are all less than 0.5% which implies good agreement. In addition, the differences are one order of magnitude smaller than the amount of backscattered radiation and about the same order to the statistical uncertainty in our simulations. Thus, the backscatter correction for photon output obtained in this study agrees well with those derived from Liu *et al.*[10]

CONCLUSIONS AND DISCUSSION

The 10 MV photon fields were modeled using a Monte Carlo technique. The optimal parameters of the incident electron beam on the x-ray target were estimated by matching the central percent depth dose curve and the dose profiles of the largest field size used (30 x 30 cm^2), with the measured data. The best estimated beam energy was found to be 10.4 MeV with a FWHM of 1.0 mm. Dosimetric data for other field sizes were also compared between the calculated and measured values and were found to be in good agreement, validating the optimal parameters. Consequently, the photon outputs were calculated for the field sizes of 5 x 5, 7 x 7, 10 x 10, 15 x 15, 20 x 20 and 30 x 30 cm^2. The discrepancies from the measured photon output were assumed to be due to the backscattered radiation from the collimator jaws into the monitor chamber. Therefore, in this study, the ratio between the measured and the calculated photon output was considered to be the backscattered correction for photon output. On the other hand, the backscattered corrections for each field sizes were also computed using the relative backscattered dose given by the formula described in Liu *et al.*,[10] where the agreement was better than 0.5% in all six evaluated cases.

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The results of our study support not only the simulation of the 10 MV photon beam from the Clinac 2100C medical linear accelerator, but also the formula for the determination of the correction factor for photon output resulted from the backscattered radiation into monitor chamber of the linac head reported by Liu *et al.*[10] Since the beam delivery is decided by the monitor unit using the dose in the monitor chamber, the dose per monitor unit will be affected by the backscattered radiation into the monitor chambers. The smaller field sizes had a larger proportion of backscattered radiation than the larger field sizes, since the main contribution of the backscatter comes from the upper collimator jaws. To improve the accuracy of the dose calculation by the monitor units, including the output factor, the effect of backscatter from the collimator jaws into the monitor chamber should be accounted for. This study has demonstrated the accountability of the formula in the case of the square symmetric field sizes ranging from 5 x 5 to 30 x 30 cm^2.

One possible way to avoid having to use such a correction factor in the Monte Carlo dose calculation is to score both the forward and backward dose in the monitor chamber at the same time the dose in the water phantom is determined.[17] However, the simulation would have to be done for every field component, reducing its benefit and logistical applicability. Using the formula proposed by Liu *et al.*[10] is perhaps the most convenient because the computation time is minimal and the formula can be applied for any arbitrary field size of less than 40 x 40 cm^2, including various collimator jaws' positions.

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