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An investigation of the depth dose in the build-up region, and surface dose for a 6-MV therapeutic photon beam: Monte Carlo simulation and measurements

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The percentage depth dose in the build-up region and the surface dose for the 6-MV photon beam from a Varian Clinac 23EX medical linear accelerator was investigated for square field sizes of 5×5, 10×10, 15×15 and 20×20 cm^2 using the EGS4nrc Monte Carlo (MC) simulation package. The depth dose was found to change rapidly in the build-up region, and the percentage surface dose increased proportionally with the field size from approximately 10% to 30%. The measurements were also taken using four common detectors: TLD chips, PFD dosimeter, parallel-plate and cylindrical ionization chamber, and compared with MC simulated data, which served as the gold standard in our study. The surface doses obtained from each detector were derived from the extrapolation of the measured depth doses near the surface and were all found to be higher than that of the MC simulation. The lowest and highest over-responses in the surface dose measurement were found with the TLD chip and the CC13 cylindrical ionization chamber, respectively. Increasing the field size increased the percentage surface dose almost linearly in the various dosimeters and also in the MC simulation. Interestingly, the use of the CC13 ionization chamber eliminates the high gradient feature of the depth dose near the surface. The correction factors for the measured surface dose from each dosimeter for square field sizes of between 5×5 and 20×20 cm^2 are introduced.

Keywords: dose in build-up region; surface dose; Monte Carlo simulation; correction factor

INTRODUCTION

One choice of cancer therapy is the use of a localized high-energy photon beam focused in the region of the tumor. As the beam passes through the patient, it interacts with tissues forming highly reactive radicals in the intracellular material, and denatured cellular components that can cause lethal damage to the irradiated cells. The deposited energy in the tissue from the irradiation per unit mass of tissue is known as the radiation absorbed dose. The higher the absorbed dose the greater is the chance of killing cells.

The dose accumulated at the boundary between the air and the patient's skin is known as the surface dose. For radiotherapeutic irradiation, it is about 75–95% of the maximum dose for an electron beam and about 10–30% for a photon beam [1]. The dose deposited within the first few millimeters of skin depth varies considerably due to the build-up character of the photon beam. To avoid skin complications, the amount of surface dose is typically taken into account simultaneously with the treatment plan. Knowing the accurate surface dose is, therefore, essential for assessing the skin damage, guiding the bolus thickness decision, and designing both the treatment technique and the scheme of dose fractionation.

Normally, the surface dose from a therapeutic photon beam is primarily contributed by the secondary charged particles, such as electrons and positrons, arising from the photon interactions in the air, in the collimator, and in the scattering material in the path of the beam [2, 3]. Therefore, the surface dose depends on the patient-specific parameters, such as the beam energy, field size, angle of beam incidence, air gap and beam-modifying devices. Previous studies have shown that the surface dose decreases relative to the maximum dose with increasing photon beam energy, while it increases with a increasing field size [4–6]. Moreover, if the location of the dose measurement lies within the build-up region, then perturbation may occur from contaminated electrons in the beam and secondary electrons generated in the irradiated medium. These electrons drop off rapidly with depth due to the high density of interactions with the atomic electrons in the medium.

In the past, the surface dose from therapeutic photon beams was investigated using one of various types of dosimeter, such as a thermoluminescense dosimeter [7–9], radiochromic film [10–13], and many types of parallel-plate ionization chambers [14–20]. Because the conditions required for the equilibrium of the charged particles are not met in the build-up region, and the dose gradient is also high, the use of an extrapolation chamber is suggested for a more reliable measurement of the dose in this region [14–16]. An extrapolation chamber is one category of parallel-plate ionization chamber, with a small sensitive volume which can be varied. Its response is reported to be very good in the non-electronic equilibrium region [21]. Therefore, the surface dose can be estimated by measuring the ionization per unit volume as a function of electrode spacing, and then extrapolating the data to a zero electrode spacing. Unfortunately, the use of extrapolation chambers is limited since they are not typically available in most institutes. Also, its measuring procedure is time-consuming which makes accurate measurement of the surface dose in a clinical setting impractical.

In contrast, fixed-electrode separation chambers are commonly available and also convenient to use for the surface dose measurement in clinical situations. However, their accuracy in the build-up region remains in doubt since there exists a cavity perturbation from the chamber volume that causes excess ionizations. To obtain an accurate surface dose value, the ionization reading has to be corrected by taking into account the perturbation conditions. Velkley *et al*. [22] proposed the correction factors derived from aluminium-walled extrapolation chamber measurements obtained by adjusting the depth dose curves in the build-up region from several types of fixed parallel-plate ionization chambers. However, Nilsson and Montelius [14] showed that the application of those correction factors to different chamber structures led to a significant error in the derived surface dose value. By evaluating the magnitude of the over-response dose at the surface and in the build-up region obtained with four different commercial parallel-plate chambers, Gerbi and Khan [16] modified the correction factor introduced by Velkley *et al*. [22] to take into account the effect of the collector edge–sidewall distance of the chamber, and to correct for the chamber response for different depths and different beam energies. Moreover, Bjarngard *et al*. [23] described an experimental method to determine the dose near the surface under lateral equilibrium conditions using a mathematical extrapolation based on the Monte Carlo (MC) simulation-calculated kerma values.

The MC method is generally considered to be an accurate tool for dose estimation in radiotherapy since the beam's particles are tracked individually in the media according to the reliable interaction database. Although an undesirable discrepancy between measurements and MC simulation-derived calculations have been reported for the initial build-up region, several recent studies have demonstrated the coherence between the results from MC simulations and the measurements obtained using an extrapolation chamber [13, 19, 24]. Devic *et al*. [13] investigated the deposited dose within the first millimeter of the build-up region and the last millimeter of the build-down region for a 6-MV photon beam using several dosimeter types and MC simulations, while Parsai *et al*. [19] studied the variation in the percentage depth dose in the build-up region using MC simulations for 6- and 10-MV photon beams from two commercial accelerators in comparison with the derived data. Both studies showed excellent agreement between the measurements using the extrapolation chamber and the MC simulations.

In this study, the central-axis dose in the build-up region and the surface dose of a 6-MV therapeutic photon beam with four different square field sizes (5×5, 10×10, 15×15 and 20×20 cm^2) from a Varian Clinac 23EX linear accelerator were evaluated using MC simulations. The dose measurements were also obtained empirically using four different commonly used dosimeters: a thermoluminescense dosimeter (TLD), a p-type photon semiconductor dosimeter, a cylindrical ionization chamber, and a parallel-plate ionization chamber. The simulated results were analyzed and compared with the actual measured data in order to establish a correlation. Finally, based on the MC simulation data, the correction factors for the measured surface dose obtained from each detector are presented as a function of the length of the square field's side.

MATERIALS AND METHODS

Monte Carlo simulation

The MC simulations were based on the EGSnrc code system, developed by the National Research Council of Canada (NRC) [25, 26]. The 6-MV photon beam generated from a Varian Clinac 23EX medical linear accelerator was simulated using the BEAMnrc user code. The dose distribution in a phantom was obtained by the use of the DOSXYZnrc user code.

The linac's head components of the Varian Clinac 23EX were built individually in the BEAMnrc user code in terms of the component module (CM). The SLABS CM was used for the target, CONS3R for a primary collimator, SLABS for a vacuum window, FLATFILT for a flattening filter, CHAMBER for an ionization chamber, MIRROR for a mirror, JAWS for a secondary collimator, and DYNVMLC for a multileaf collimator. The description of the dimensions, geometries and materials of all components were taken from the manufacturer's detailed specifications.

Since the actual information about the incident electron beam on the target inside the linac was experimentally unknown in our study, the energy and the Gaussian distribution of the radius of the electron beam were determined from the best match between the simulated and measured results for the percentage depth dose along the central axis and the beam profiles at a 10-cm depth for photon beam field sizes of 10×10 and 30×30 cm^2. The cylindrical ionization chamber of type CC13 was used to measure the depth dose in a water phantom. Since the measured dose in the build-up region using this detector is uncertain, our matching condition on the percentage depth dose started from the depth at a maximum dose to 30 cm. The obtained monoenergetic electron energy and the Gaussian distribution radius tuning were 6.1 MeV and 1.2 mm, respectively. In the simulation, the transport parameters used in the BEAMnrc user code for generating the phase-space files and in the DOSXYZnrc user code for calculating the deposited dose in the medium were as follows: ECUT = 700 keV, PCUT = 10 keV, AE = 700 keV and AP = 10 keV. The particle's transport is terminated and its residual energy is transferred in the current region when the total energies of the electron and photon are less than the value of ECUT and PCUT, respectively. The production of secondary particles is considered if the particle's total energy is greater than AE for the knock-on electrons and greater than AP for the bremsstrahlung photons. To specifically calculate the surface dose, both the ECUT and AE values were lowered to 521 keV. The values of the transport parameters were selected to ensure the accuracy of the computed dose to be better than 1% [27]. Additional details on the transport parameters can be found in the BEAMnrc and DOSXYZnrc user manuals [28–30]. The phase-space file, which contains the information on each beam's particle, was recorded on the plane perpendicular to the beam-axis at a 90 cm source-to-surface distance (SSD). The number of initial electrons incident on the target was set to be about 10^9 events and the average scored particle in each phase space file was about 5×10^7 events. The efficiency of our simulations was also enhanced by the use of the variance reduction technique [27].

Using the phase space file obtained from the BEAMnrc code as the input for the DOSXYZnrc code, the beam's interaction in the $30 \times 30 \times 20$ cm^3 water phantom was simulated and, hence, the deposited dose within a voxel was obtained. For determination of the build-up dose, the voxel size along the beam central axis from the phantom's surface to a few millimeters depth was set to $1 \times 1 \times 0.014$ cm^3 for the 5×5 cm^2 field size and $3 \times 3 \times 0.014$ cm^3 for the three other larger square field sizes (10×10, 15×15 and 20×20 cm^2). The number of simulation events gave a statistical uncertainty of less than 1% for the deposited dose in the smallest voxel size.

Measurements

The Varian Clinac 23EX linear accelerator, equipped with a Millennium 120-leaf MLC and on-board imaging system, located at the Department of Radiation Oncology, Siriraj Hospital, Thailand, was used in this study. All measurements were performed on the 6-MV photon beam along the central axis with square open field sizes of 5×5, 10×10, 15×15 and 20×20 cm^2 at a constant source-to-surface distance of 100 cm. The complete percentage depth doses were measured in a Blue water phantom (Wellhofer Scanditronix, Germany) at a depth ranging from 0 to 30 cm with a scanning resolution of 2 mm. The detectors used with this water phantom were a compact cylindrical ionization chamber of type CC13 (Wellhofer Scanditronix, Germany) and a silicon p-type photon semiconductor dosimeter of type PFD (Wellhofer Scanditronix, Germany). The CC13 dosimeter has an active diameter of 6 mm, a cavity volume of 0.13 cm^3, and a wall thickness of 0.07 g/cm^2. The PFD dosimeter has an active diameter of 2 mm and an effective thickness of 0.06 mm from the detector's front surface. These two dosimeters are routinely used to acquire the common beam data, such as the percentage depth dose, the beam profiles and the output factor.

We also obtained the central axis depth dose using the parallel-plate ionization chamber (Markus 23392, PTW-Freiburg) and the TLD (HARSHAW Chemical Co, Solon, OH). The measurements were performed in $30 \times 30 \times 20$ cm^3 solid water equivalent phantom slabs (RMI, Model-457) at a depth of 0, 0.2, 0.3, 0.5, 1, 1.2, 1.5, 2 and 3 cm. The solid water phantom has a physical density of 1.04 g/cm^3. The readings from each measured position were corrected and then normalized to the maximum depth dose. The effective measured point of the Markus chamber was assumed to be at the bottom of the entrance window electrode. The plate separation of this detector is 2 mm, with a 0.35 mm distance between the side wall and the collector. Each measured signal was taken from an average of five readings from an output variation acquired by the DOSE1 Electrometer (iba dosimetry). In order to take into account the polarity effects, the measurements with both positive and negative voltage (+300 V and –300 V) were performed and the polarity correction factor was found to be 0.98.

The charge obtained from the fixed-separation parallel-plate ionization chamber in the build-up region was mainly contributed from electrons scattered from the sidewalls of the chamber and collected in the chamber active volume. The correction factors for this perturbation effect were first introduced by Velkley *et al.* [22], based on extrapolation chamber measurements at a percentage depth dose in the build-up region, as measured by a fixed size parallel-plate chamber. The Velkley formula has been used in several studies and was originally intended to apply to all types of fixed-separation parallel-plate chambers. Nevertheless, Gerbi and Khan *et al*. [16] demonstrated that the Velkley formula, obtained with only one extrapolation chamber for a specific parallel-plate chamber, did not produce an accurate dose for all parallel-plate chambers, but rather overestimated the amount of scattering of higher energy beams. They improved the previous correction method by including (i) the effect of the collector edge-sidewall distance to the chamber, and (ii) the dose response of the chamber at different depths and different beam energies for various detector types. Here, we refer to the correction method introduced by Gerbi and Khan *et al*. [16] as the GK method. By comparing the readings of the various detectors with the reading of an extrapolation chamber, the correction equation was given by Eqs. (1) and (2):

$$\mathrm{P}'(\mathrm{d}) = \mathrm{P}(\mathrm{d}) - \xi(E,0)\ell\exp(-\alpha(\mathrm{d}/\mathrm{d}_{\max})) \tag{1}$$

$$\xi(E,0) = 27.19 - 32.59\ \mathrm{IR} + (-1.666 + 1.982\ \mathrm{IR})\mathrm{C} \tag{2}$$

where P′(d) and P(d) are the corrected and uncorrected percentage depth dose at depth ‘d’, respectively, E is the maximum energy of the photon spectrum, ℓ is the plate separation in the units of mm, ξ(E, 0) is the over-response in percent per mm of chamber plate separation at the phantom surface, IR is the ionization ratio measured at 10 and 20 cm depths for a field size of 10×10 cm^2 at 100-cm SSD, C is the collector edge to side wall distance in mm (it is 0.35 in our case) and α is an empirically determined constant of proportionality which is equal to 5.5. The following constants were determined for our PDD measurement from the 6-MV photon beam: IR = 0.6709, ξ(E, 0) = 5.26% per mm, and d_{max} = 1.5 cm.

Rawlinson *et al*. [31] then investigated the design features of a fixed-separation parallel-plate ionization chamber that has a negligible chamber signal of electrons from the side walls, and developed a guideline for predicting this sidewall effect. The formula was introduced to estimate the over-response for a commercial ionization chamber under a normal build-up condition, and differs from the GK method in the parameter of the guard width that is used to characterize the proximity of the sidewall. However, this correction agrees well with the results of GK method for small collector diameters.

The TLDs used in this dose measurement are a lithium fluoride (LiF) crystal doped with magnesium and titanium (HARSHAW Chemical Co, Solon, OH). Their thickness is 0.39 mm with a 9.92 mm^2 surface area (3.15×3.15 mm^2). The effective point of measurement for the TLD was assumed to be at the middle of its thickness. The TLD chips were carefully placed on the solid water phantom and irradiated at a depth of 0, 0.2, 0.3, 0.5, 1, 1.2, 2 and 3 cm. Three repeated measurements were made to obtain the dose at each position. Before irradiation, all TLD chips were placed in an annealing oven at 400°C for 1 h followed by 100°C for 2 h. Then, they were exposed to a known dose of 1 Gy at the depth of maximum dose from a 15×15 cm^2 field of a Cobalt-60 machine (THERATONIC 780C). The dose received by the TLD chip was recorded by the TL reader (Model 5500; HARSHAW Chemical Company, Salon, OH). As a result, the correction factors for each individual TLD chip were obtained. The process was repeated three times to produce a reliable correction factor. The variation of the TLD response with respect to the changes in either the field size or the measurement location was not included in this study, since the effect of photon spectral variations on the response was less than 1% for all of our dose measurements [32].

RESULTS

Monte Carlo simulations

The depth doses along the central axis of the beam were obtained for different square field sizes of our 6-MV photon beam simulation using the DOSXYZnrc code. They were normalized as a percentage of the maximum dose. Here, the percentage dose near or at the phantom surface was estimated by the third-order polynomial extrapolation of the simulated data. The observed percentage doses at the surface and at a depth of 0.0007 and 0.005 mm were compared with the published results of Devic *et al*. [13] and Parsai *et al*. [19], since they had performed reliable measurements of the surface dose and the dose in the build-up region at the same beam energy from similar medical linear accerelators (the Varian Clinac 1800 and the Varian 2300 (C/D), respectively). Accordingly, a substantial discrepancy in the dose in the build-up region from our MC simulation-based calculations and their previously reported empirical measurements was not anticipated. Table 1 summarizes the dose comparison using the MC simulation results obtained here and the previously measured values [13, 19] for the square fields with lengths of 5, 10 and 15 cm. Consistent results were generally observed in which all of the differences were less than 2%. Therefore, we conclude that our calculated surface doses using the MC simulation were justified.

Measurements of percentage depth dose

The percentage depth doses along the beam central axis for our 6-MV photon beams were acquired experimentally with four detectors for the four square field sizes of 5×5, 10×10, 15×15 and 20×20 cm^2. The readings from each detector were assigned to the effective point of measurement for each individual detector. For the CC13 chamber, the effective point was determined automatically by the

Table 1: Comparison of the percentage doses at the surface, and at a depth of 0.0007 and 0.005 mm for the 6-MV photon beams obtained from our MC simulation and from previously reported empirical measurements [13, 19]

Depth (mm)	Field size (cm^2)	Measured value [13, 19]	MC simulation	Difference
0	5 × 5	10.53	10.27	–0.26
	10 × 10	16.04	16.45	+0.41
	15 × 15	21.74	22.22	–0.48
0.0007	5 × 5	11.50	11.61	+0.11
	10 × 10	17.00	18.30	+1.30
	15 × 15	23.60	23.61	+0.01
0.005	5 × 5	28.32	28.12	–0.20
	10 × 10	33.34	33.48	+ 0.14
	15 × 15	38.02	37.35	–0.67

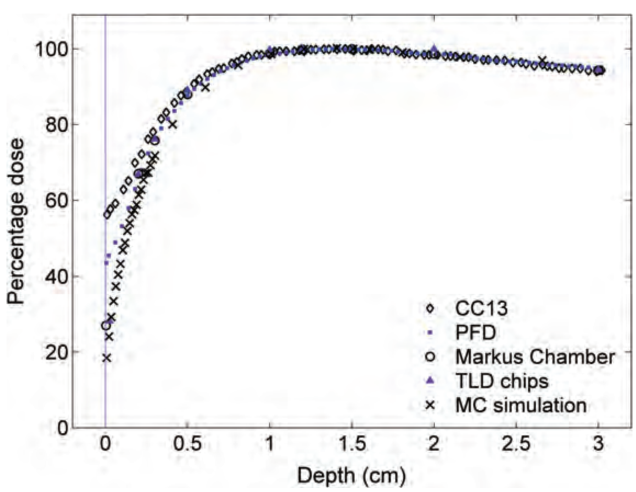

Figure 1: The percentage depth dose curves obtained using the CC13 chamber, PFD dosimeter, Markus chamber, TLD chips and the Monte Carlo simulation, for the 6-MV photon beam with a 10 × 10 cm^2 field.

computerized scanning system. The effective point of measurement for the PFD dosimeter and Markus chamber was assumed to be at the front surface and at the bottom of the entrance window electrode, respectively. For the TLD chip, the effective point was assigned to the middle of its thickness and scaled by its density of 0.0496 g/cm^2.

The depth dose curve comparison for the measurement and the simulation for a 10 × 10 cm^2 field is shown as an example in Fig. 1. For all field sizes, we found excellent agreement between all measured and simulated percentage depth doses for the depth beyond the build-up region or the depth after the maximum dose. However, near the surface a large deviation from the MC simulation result is noticed between the measurement using the CC13 chamber and the PFD dosimeter, while the results from the TLD and the Markus chamber seem to be in good agreement with the MC simulation. This suggests that reliable measurement of the percentage depth dose beyond the build-up region can be obtained from any of our four detectors. However, the dose measurement in the build-up region is strongly dependent on the detector.

Focusing on the near-surface region, we examined the percentage depth doses obtained from the four detectors as well as from the MC simulation for a 6-MV photon beam with the four different square field sizes (5 × 5, 10 × 10, 15 × 15 and 20 × 20 cm^2). The trendlines for each data set were obtained from the least square fitting using a polynomial function of an appropriate order. A similar change was observed in the near-surface region between the measured data from both the Markus chamber and the TLD chip, and the MC simulated data was observed (Fig. 2), in which these curves have a negative curvature. In contrast, the CC13 and PFD data show a positive curvature. The dose at zero depth measured by each detector was always higher than that of the MC simulation for all four evaluated square field sizes, consistent with previous studies [13, 14, 16]. Interestingly, the dose readings from both the CC13 chamber and PFD dosimeter are almost steady near the surface, varying less than 5% for the CC13 chamber and 10% for the PFD at within 1 mm from the surface for the 5 × 5 up to 20 × 20 cm^2 field sizes, while it changes by almost 30% in the MC simulation. The insensitivity of the depth for the dose measurement near the surface using these two detectors implies that a very precise position of the measurement may not be necessary.

The dose at zero depth from the TLD measurement had the best correlation with the MC simulations, with the percentage surface doses from the TLD data being 14.6, 20.5, 25.3 and 32.2 for the field sizes of 5 × 5, 10 × 10, 15 × 15 and 20 × 20 cm^2, respectively. The worst and second worst correlation with the MC simulated data was exhibited by the CC13 and PFD detectors, respectively. These detectors had a relatively large sensitive volume, and they are also made of non-water equivalent material, which leads to an over-response dose near the surface region where a steep dose gradient exists. Therefore, the CC13 chamber and PFD dosimeter are not recommended for surface dose measurements unless their accurate correction factors are available.

Correction factor for the measured surface dose

From the extrapolation of the measured dose in the build-up region, the percentage doses at zero depth (surface doses) for the 6-MV photon beam with different field sizes are shown in Fig. 3. The measured surface dose clearly increases with increasing field size, regardless of the

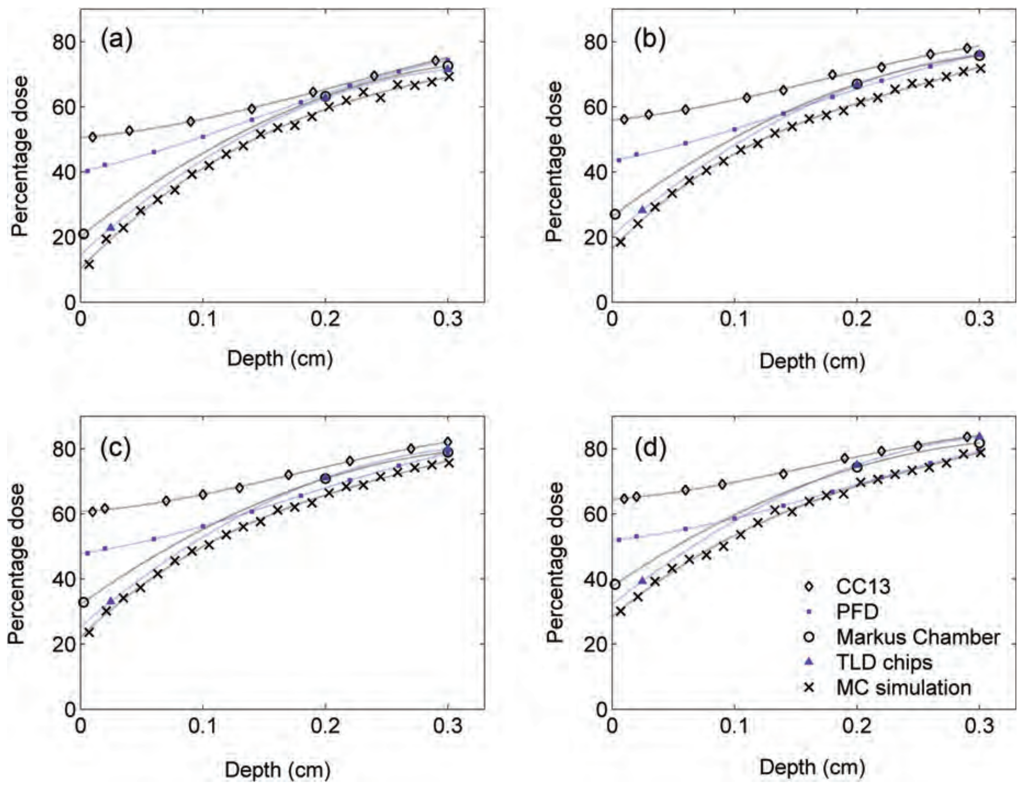

Figure 2: The percentage depth dose for the four different square field sizes of **(a)** 5×5, **(b)** 10×10, **(c)** 15×15 and **(d)** $20 \times 20\ cm^2$, obtained from the four different dosimeters, plus the MC simulation data, normalized to the maximum depth dose and the best trend line for each data set.

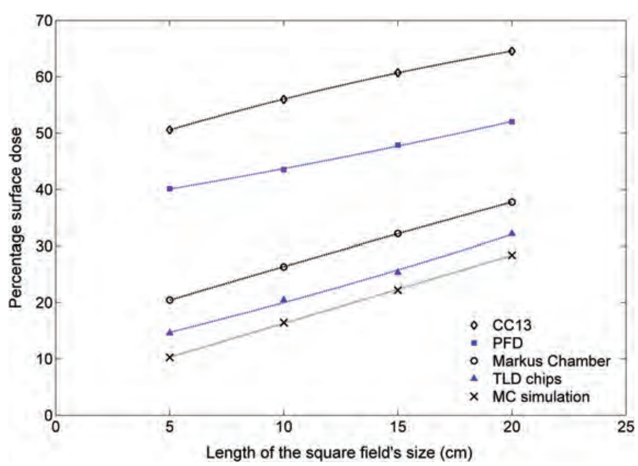

Figure 3: The percentage surface dose obtained from the four different dosimeters and the MC simulation, normalized to the maximum depth dose for each of the four square field sizes at a 100-cm SSD for the 6-MV photon beam.

detector used in the measurement, and this is also observed with the MC simulation. This is mainly due to the increasing number of scattered electrons in the air and collimator. The measurements obtained from the TLD chip and Markus chamber gave surface dose values close to that of the MC simulation, while a very large discrepancy was found when using the PFD dosimeter and, especially, the CC13 chamber. Before scaling of the effective depth of measurement, the over-response of the TLD chip and Markus chamber for the surface dose were about 10%, which is consistent with the reports of Devic *et al*. [13] and Gerbi and Khan [16]. After taking into account the effective point of measurement, the over-response from the TLD chip was reduced to approximately 4%, while that for the Markus chamber remained almost unchanged due to its larger effective volume.

The high percentage surface doses observed with the PFD and CC13 dosimeter with the smallest field size ($5 \times 5\ cm^2$) were about 40% and 50%, respectively (Fig. 3). Possibly, these two detectors pick up many low-energy electrons from the non-electronic equilibrium situation in the build-up region. Moreover, at some point of the measurement, and especially close to the phantom surface, some part of these detectors was above the water level, which is also a cause of the non-equilibrium. Note that the percentage dose near the surface obtained with both the PFD and CC13 chamber did not change rapidly with the increasing depth, but remained approximately steady near the surface phantom for all four tested square field sizes (Fig. 2). In other words, these detectors eliminate the high gradient feature of the dose in the build-up region of the photon beam. As a result, their position for measuring the surface dose is not very critical. Because of this, both the PFD dosimeter and CC13 chamber can still be considered as useful detectors for the surface dose measurement as long as their accurate correction factors are given.

In order to scale down the over-response of the measured surface dose, the correction factors for each dosimeter used in this study were evaluated by calculating the ratio of the MC simulated surface dose and the measured value. For a field size of 10×10 cm^2, the correction factor was 0.294, 0.378, 0.625 and 0.804 for the CC13 chamber, PFD, Markus chamber and TLD chips, respectively. Noting that a smaller correction factor implies a larger over-response of the detector for the surface dose measurement, the over dose response order (highest to lowest) was found to be the CC13 chamber > PFD > Markus chamber > TLD chip. The relationship between the correction factors, based on the data from the MC simulation as the reference value, for the four detectors at various open square field sizes are shown in Fig. 4. The correction factors for the measured surface dose obtained from the TLD chips are almost independent of the square field size, with an average correction factor of 0.817. The correction factors for the TLD chip are, of course, the largest of the four different types of detectors, since we found that this detector gave the nearest surface dose to the MC simulations. In order to include the effect of the field size dependence in the surface dose correction factor, we performed the least square fitting on the correction factors for the four square field sizes, assuming a parabolic tendency on the length of the field's side. The trendlines for the CC13 and PFD data were found to be almost linear (Fig. 4).

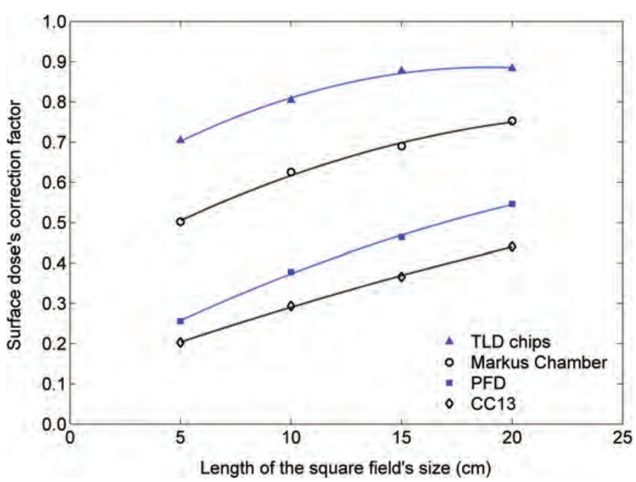

Figure 4: The correction factor for surface dose as a function of the length of square field's side for the four different dosimeters based on the Monte Carlo simulation for the 6-MV photon beam.

The correction factor (C_i) for each detector (labeled by an index i) as a function of the length of square field's side (L) can be written as in Eq. (3):

$$C_i(L) = a_i L^2 + b_i L + d_i \tag{3}$$

where a_i, b_i and d_i are the parameters which depend on the type of detector and are shown in Table 2. They are supposed to be applicable to the surface dose of the square field for a size ranging from 5×5 to 20×20 cm^2. The surface dose after the correction is simply obtained by a multiplication of the correction factor C_i with the reading surface dose from the measurement.

The surface doses for three other square field sizes (3×3, 12×12 and 15×15 cm^2) were empirically investigated using the measurements derived from the Markus chamber in a water equivalent solid phantom and, also, by MC simulation. Only the 12×12 cm^2 field is within the range of our previously studied field sizes, the other two extending this range to larger and smaller field sizes. The percentage surface doses from the actual reading, the Eq. (3) corrected values using the empirical correction factors (Table 2), and the MC simulation values are shown in Table 3, along with the modified percentage surface doses using the GK method for the Markus chamber. The corrected surface dose values (using Eq. (3) and Table 2) were lower than that of the MC simulation for the 3×3 and 12×12 cm^2 field sizes, and were closer to the MC simulation than the GK corrected values for only the 12×12 cm^2 field size. For both the 3×3 and 15×15 cm^2 field sizes, the GK corrected values were closer to the MC. In other words, the empirical correction of the surface dose presented here (Eq. (3) and Table 2) is not better than the GK method outside the range of the studied field size (5×5 cm^2 to 20×20 cm^2). Therefore, these empirical correction factors C_i should only be used for square field sizes in the range of between 5×5 to 20×20 cm^2.

Table 2: Fitting parameters for the correction factor of the surface dose defined in Eq. (3) for the four different types of dosimeters

Type of dosimeter	Coefficient[a]		
	a	b	c
CC13	–0.0002	0.0198	0.1091
PFD	–0.0004	0.0291	0.1220
PTW model 23392 (Markus)	–0.0006	0.0314	0.3628
TLD chip	–0.0009	0.0357	0.5470

[a]See Eq. (3) in the text.

Table 3: The percentage surface doses for a 6-MV photon beam with a square field size of 3×3, 12×12 and 25×25 cm^2, obtained from measurements using the Markus chamber and then corrected using the GK method or empirical correction factor C_i (Table 2 and Eq. (3)), in comparison with that from MC simulations

Field size (cm^2)	Measurement (nC)	Percentage surface dose (%)		
		MC simulation	GK method	Our method
3×3	18.62	9.71	8.21	8.14
12×12	29.38	20.03	18.97	19.19
25×25	43.28	30.97	32.88	33.45

DISCUSSION

The MC simulated depth doses near the surface for this 6-MV photon beam were all in good agreement with the previously reported measurements from similar machines using an extrapolation chamber. For the beam used here, we measured the percentage depth doses along the beam's central axis for four square field sizes (5×5, 10×10, 15×15 and 20×20 cm^2) using four different detectors (TLD chips, PFD dosimeter, CC13 and Markus chamber). The consistency between the measured data from all four detector types and the MC simulated data were observed for depths beyond the depth of the maximum dose, but were all clearly different from the MC simulated data in the build-up region. Compared to the MC simulated data, the measured doses were typically overestimated near the surface region. The surface doses from each dosimeter were then estimated by extrapolation of the measured doses near the surface, and were found to increase with increasing field sizes for all four types of dosimeters. For the CC13 chamber data, we found that the measured doses varied less than 5% within 1 mm from the surface for all four evaluated square field sizes, which means that its accurate positioning for surface dose measurement is not critical. From the MC simulation, the simulated dose can vary up to 30% from the surface to 1-mm depth in the phantom. As a result, the CC13 chamber is likely to be suitable for reliable surface dose measurement only if its over-response is accurately known. Finally, we have derived the correction factors for the surface dose measured by each of the four types of detectors as a function of the length of the square field's side ranging from 5 to 20 cm. These are always less than 1.0 due to the over-response of the detectors. The smallest correction factor, or the largest over-response, was found with the CC13 chamber, whilst the largest correction factor, or the smallest over-response, was found with the TLD chip.

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