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Independent absolute dose calculation using the Monte Carlo method on CT-based data

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Original Article

Abstract

Purpose: The accuracy of delivered dose is essential to the quality of radiotherapy treatment and tumor response. Generally, there are many types of dosimeter have been used to verify the dose from the treatment; however, most of these dosimeters are impractical for clinical situation. The goal of this study was to assess an absolute dose derived from the Monte Carlo (MC) method for the so-called 6- and 10-MV photon beams obtained from Varian Clinac 2100C linear accelerator. **Methods:** The deposited doses have been calculated by the EGSnrc code system and, then, were converted into the absolute doses. We were also measured, in water phantom, by an ionization chamber and, in the chest region of Rando phantom, by a thermoluminescense dosimeter (TLD). **Results:** The simulated data in water phantom agree with the results from both the measurement and previous studies within 2%. By comparing the absolute dose at various positions within the Rando phantom from two-opposing irradiated fields, the difference from MC calculation and TLD measurement was within 2%. Unfortunately, the calculated doses obtained from the collapse cone convolution (CCC) algorithm showed notable difference from that of the MC method. For the interface region within the provided field, it was higher than that from the MC method by almost 5% for the 6-MV and 7% for the 10-MV photon beam. **Conclusion:** Our findings indicated that the MC method was on the level with the measurement for the dose determination, especially within the delivered field to a heterogeneous phantom.

Keywords: Monte Carlo method, Absolute dose, TLD measurement, Rando phantom

1. Introduction

Monte Carlo (MC) methods have many fields of application in radiation therapy and are recognized as the alternative tool for simulating beam passing through heterogeneities region particularly for lung and bony anatomy, and tissue interfaces. However, the clinical implementation of these techniques has been limited by long calculation times. Consequently, MC simulations have only been used as an important quality assurance (QA) tool for relative comparison with measurement or calculation using dosimeters and commercial treatment planning system (TPSs), respectively.

Because of the limitation of measurements in real patient-specific, MC method offers an alternative determination of dose delivered to the entire treatment volume in real patient geometry and heterogeneities. Although TPS can give dose distribution conveniently, the significant differences between MC and TPS calculation in such complex situations are still found[1-4]. For example, Han *et al.*[1] previously compared various dose calculation algorithms for homogeneous water and multilayer slab virtual phantoms. The study reported large dose deviation for the collapse cone convolution (CCC) algorithm in the bone, lung, and interface regions

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in which the spatial distributions of these differences depend on the field sizes and energies.
Since the clinical treatment plan is commonly presented in term of absolute dose on patient computed tomography (CT) data sets, the accuracy of absolute dosimetry in MC simulation should be concerned. Several reports propose the algorithm for absolute dosimetry based on MC data[4-8], however no benchmark information is provided regarding the accuracy of the MC-calculated absorbed doses. The calculation from the studies of Francescon *et al.*[5], and Leal *et al.*[7] did not include the backscatter into the monitor chamber. In contrast, Popescu *et al.*[4] have provided the absolute dose formula accumulated the backscatter from the jaws entering the monitor chambers. The MC-calculated dose and the experimentally absorbed dose were compared to verify the MC calculation. According to their investigation, the MC absolute dose calculations normalized by the incident particle are in excellent agreement with experiment with the percentage differences of less than 2%.

The purpose of this study was to investigate the absolute dose calculation performance of MC method on the CT-datasets for both 6- and 10-MV therapeutic photon beams obtained from a Varian Clinac 2100 C linear accelerator. The simulated results derived in the water phantom and the Rando phantom were analyzed and compared against the CCC dose calculation, ionization chamber and TLDs measurements.

2. Methods and Materials

2.1. Measurements

Varian Clinac 2100C (Varian, Palo Alto, USA) linear accelerator was used in this study. The measurements were performed at the presumed 6- and 10-MV photon beam. In this study, the percentage depth doses (PDDs) and the beam profiles for the 20×20 cm^2 field size were obtained to investigate the performance of our MC simulation. The PDD curves and the beam profiles were acquired using the RFA-300 dosimetry system (Wellhofer Scanditronix GmbH, Germany) at the depth ranging from 0 to 30 cm. The dose scanning step was 2 mm using a silicon p-type photon semiconductor (Wellhofer Scanditronix, Germany) dosimeter.

To ensure the reliability of our simulation for the other field sizes, we also obtained the actual measured dose in a water phantom (Wellhofer Scanditronix, Germany) using a Farmer ionization chamber type FC65-G (Wellhofer Scanditronix, Germany) and a parallel-plate ionization chamber type 34001 (PTW-Freiburg) connected to the DOSE-1 Electrometer (iba dosimetry). Commonly a Farmer chamber and a parallel-plate chamber is the standard dosimeter for clinical use to acquire the machine output and the absolute dose at shallow depth, respectively. Measurements were performed by a Farmer chamber along the central axis of the radiation beam for five square field sizes of 5, 10, 15, 20 and 30 cm^2 with a constant source-axis distance (SAD) of 100 cm at the depth of 5, 10 and 20 cm. While a parallel-plate chamber was used to acquire the measured data in the same situation at the maximum depth to avoid the effect of electron contamination.[9] Each measured signal was taken from an average of five readings for an output variation. The absorbed dose was determined and calculated followed the IAEA TRS-398 dosimetry protocol.[9]

Other measurements were performed using thermoluminescense dosimeters (HARSHAW Chemical Co, Solon, OH) in the chest region of Alderson radiation therapy phantom (Model 457; Radiology Support Devices, USA) to investigate the heterogeneity effect. Lithium fluoride (LiF) crystals doped with magnesium and titanium in the form of TLD rods (HARSHAW Chemical Co, Solon, OH) was used for measuring. TLD rods have been inserted in the Rando phantom and irradiated with two-opposing fields for the mediastinal treatment. Because there was no significant difference of beam quality for LiF-TLD[10], the correction factors for each individual TLD were provided by an irradiation with a known dose from a Cobalt-60 machine (THERATONIC 780C) at the depth of maximum dose. To ensure the reading consistency, these reading from each measured position in three repeated times were acquired by TL reader (Model 5500; HARSHAW Chemical Company, Salon, OH).

2.2. Monte Carlo simulation

EGSnrc Monte Carlo code system, provided by the National Research Council of Canada[11-12] was used to simulate photon beam from the medical linear accelerator. It composes of the two sub-codes; BEAMnrc and DOSXYZnrc. The BEAMnrc code was used to model the linac's head as a series of component modules. In order to eliminate the forward dose from backscatter dose for every simulated field, the simulation of our linac head was separated. The first phase space, A, was scored above the jaws resulting in the dose accumulated in the monitor chamber due to the beam entering the chamber from above. Another phase space, B, was scored under the jaws resulting in the dose accumulated in the monitor chamber due to the particles backscattered from the jaws. DOSXYZnrc allows the radiation transport and dose deposition in the virtual phantoms or in CT data to be calculated in Cartesian coordinates. Generally, the MC dose reports the dose-to-medium in medium ($D_{(m,m)}$).

Before estimating the MC absolute doses, the initial energy and the radius distribution of the incident electron beam interacted on the target were adjusted according to the match between the simulated and measured results of the percentage depth doses and the dose profiles at a 10 cm depth for a 20×20 cm^2 field. For the depth doses, the matching condition for both 6- and

10-MV therapeutic photon beams started from the depth at a maximum dose to 25 cm. For the 6-MV photon beam, from the best match, the obtained initial energy and radius of electron beam were 6.2 MeV and 1.0 mm, respectively. The optimal incident electron energy and beam radius for the 10-MV photon beam were 10.4 MeV and 1.3 mm, respectively. These simulated parameters gave the best similar characteristics to that of the realistic photon beams.

The EGSnrc settings used in the calculation were following: the global electron transport cut-off energy (ECUT) = 700 keV, the global photon transport cut-off energy (PCUT) = 10 keV. Cross section data were generated using PEGS4. The description of other parameters could be found in BEAMnrc and DOSXYZnrc manual.[13-15] For calculating the PDD, the beam profile, and the machine output in the DOSXYZnrc code, the simulated beam interacted on a water phantom with the voxel sizes of 0.5×0.5×0.5 cm^3. The simulated dose is a mean value in each calculation voxel per incident particle of the radiation source; therefore it is proportional to the dose per monitor unit.

The results obtained in the water phantom were used to verify the accuracy of MC-based data, and to calculate absolute dose values in the CT phantom. For dose calculation in the CT-based phantom, the voxel sizes were set to 0.4×0.4×0.5 cm^3 to match that on the dose grid resolution of TPS. The simulated results were converted to the absolute dose according to the method proposed by Popsecu *et al.*[4]. The absolute dose at the point (x, y, z) in the phantom is given by equations (1) and (2):

$$D_{xyz,abs} = D_{xyz} \frac{\left(D_{ch}^{forward} + D_{ch\ (10\times10)}^{back}\right)}{D_{ch}^{forward} + D_{ch}^{back}} \frac{D_{xyz.abs}^{cal}}{D_{xyz}^{cal}} U \quad (1)$$

$$U = \frac{D_{ch} N_e}{D_{ch,abs}^{1MU}} \quad (2)$$

where D_{xyz} is the dose per incident history along the beam central axis deposited in a prescribed voxel, D_{ch} is the dose per incident history accumulated in the monitor chamber, $D_{ch}^{forward}$ is the dose contribution from the beam interaction with monitor chamber, D_{ch}^{back} is the dose contribution from the beam interaction with collimator, D_{xyz}^{cal} is the normalized dose at the calibration point of a clinical accelerator, $D_{xyz,abs}^{cal}$ is the absolute dose at the calibrated point (x, y, z) in the phantom, $D_{ch,abs}^{1MU}$ is the absolute dose at the monitor chamber corresponding to one monitor unit, U is the number of monitor units, N_e is the number of incident electrons. Table 1 summarized the following values determined from both 6- and 10-MV photon beams.

2.3. Clinical geometry

CT image datasets were acquired for a Rando phantom through a MX IDT8000 (Philips Healthcare, Andover, MA) CT machine with a slice thickness of 3 mm. During CT scanning, all holes are plugged with bone-, soft tissue-, and lung-equivalent pins. The images were acquired and imported into TPSs and DOSXYZnrc using the CTCREATE program[15] which converts the CT data into the desired dimensions, material types, and mass densities based on a CT number to density correlation.

A clinical two-field opposing conventional plan with gantry angles 0° (anterior), 180° (posterior), 150° (left-posterior), and 330° (right-anterior) for both photon beam energies was delivered to a Rando phantom. The investigated beams were the 10×12 cm^2 field for the AP-PA direction, and the 8×15 cm^2 field for RAO-LPO direction. Each of the fields was placed onto each site of phantom to generate 100 cGy to its isocenter.

2.3. Collapse cone convolution algorithm

The Collapse cone convolution (CCC) algorithm[16] in Pinnacle 7.6C TPSs (Philips Medical Systems, Inc., Fitchburg, WI) uses convolution/superposition methods to compute TERMA convolved with energy deposition kernels and to account for the effects of tissue heterogeneities. All plans calculated with DOSXYZnrc were recalculated using the CCC algorithm. The same CT number-density conversion was used for the CCC calculations. The CCC TPS also reports the dose-to-medium in medium ($D_{(m,m)}$) as default in Pinnacle.

3. Results

3.1. Absolute dose calculation using the MC data

To verify the accuracy of simulated results for the 6- and 10-MV photon beam from Varian machine using BEAMnrc/DOSXYZnrc code, the dose at the depth of maximum dose, 5, 20 cm and at the depth of 10 cm which is the practical reference depth for calibration were compared with the measured results. The difference between the absolute dose at different depths for the square open field side of 5, 10, 15, 20, and 30 cm^2 for 6- and 10-MV photon beam in water phantom were presented in Figures 1 and 2, respectively. The solid points represented the measured dose, while the light points were the simulated dose. From the results, most of our simulated data were similar to measured data with percent differences of less than 2%.

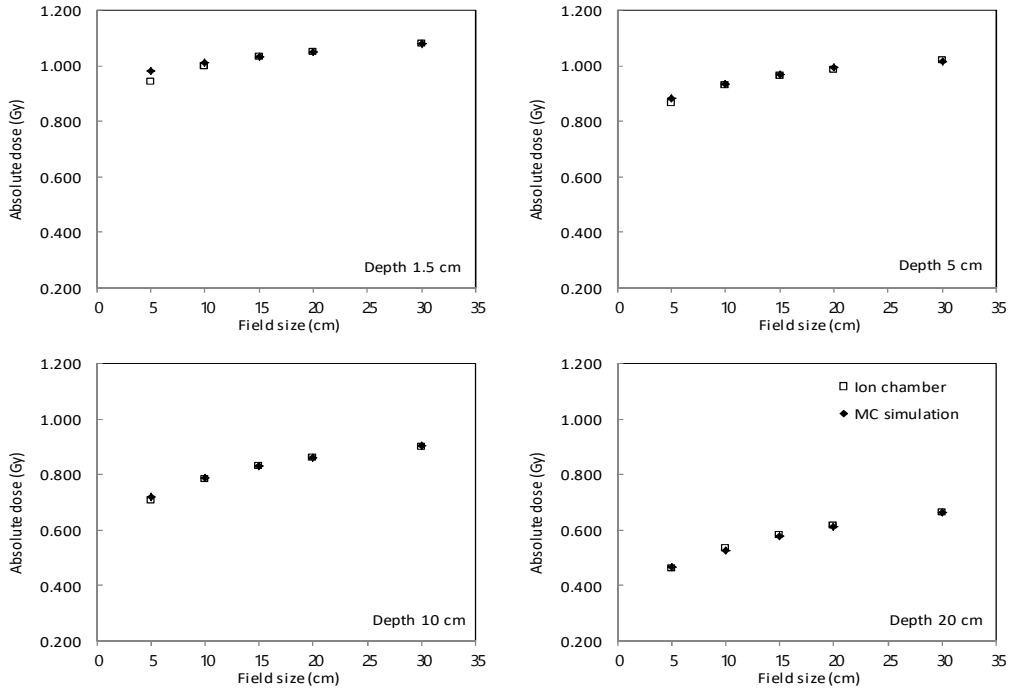

Figure 1: The absolute dose at different depths on the central axis for the square field side of 5, 10, 15, 20, and 30 cm^2 for 6-MV photon beam.

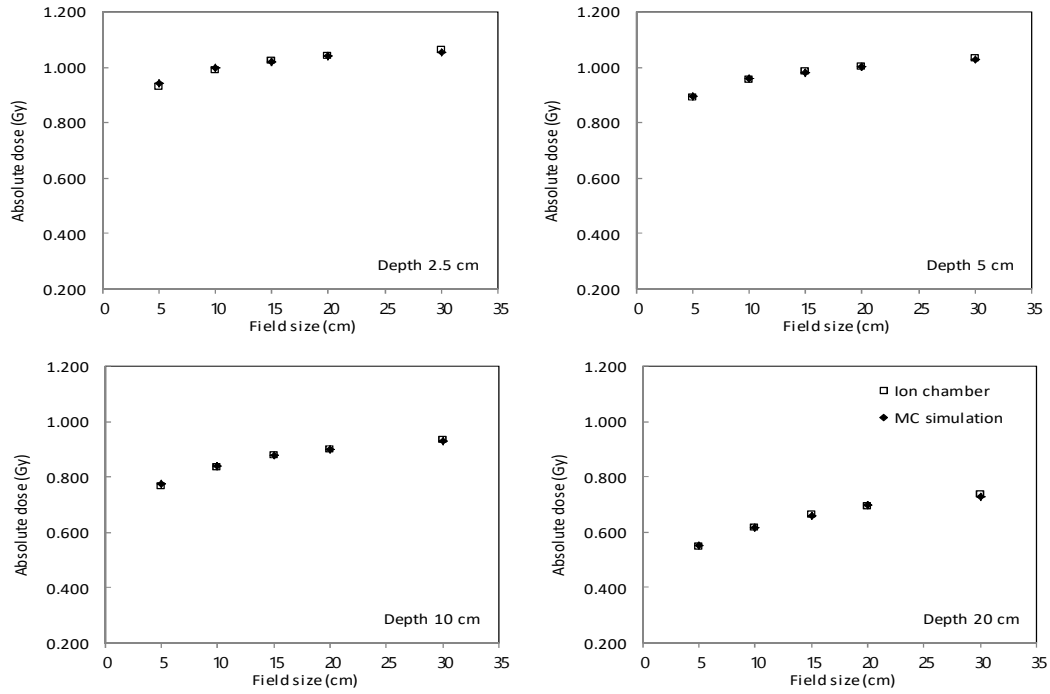

Figure 2: The absolute dose at different depths on the central axis for the square field side of 5, 10, 15, 20, and 30 cm^2 for 10-MV photon beam.

Table 1: The numerical values specific to our study for absolute dose calculation.

Parameters	Numerical values	
	6 MV	10 MV
$D_{ch}^{forward}$ (Gy/incident particle)	2.261×10^{-15} ±0.2%	5.473×10^{-15} ±0.1%
$D_{ch\ (10\times10)}^{back}$ (Gy/incident particle)	6.241×10^{-17} ±1.0%	1.748×10^{-16} ±0.8%
D_{xyz}^{cal} (Gy/incident particle)	1.066×10^{-16} ±0.9%	2.611×10^{-16} ±0.7%
$D_{xyz,abs}^{cal}$ (cGy/MU)	0.787	0.839

Table 2: Monte Carlo and measured relative output factors (ROF) for both 6- and 10-MV photon beams.

Field size (cm^2)	6 MV			10 MV		
	MC_ROF	MEAS_ROF	%Diff	MC_ROF	MEAS_ROF	%Diff
5×5	0.9123	0.8981	1.59	0.9202	0.9139	0.70
10×10	1.0000	1.0000	0.00	1.0000	1.0000	0.00
15×15	1.0559	1.0573	0.13	1.0464	1.0478	0.14
20×20	1.0940	1.0943	0.02	1.0714	1.0754	0.37
30×30	1.1499	1.1452	0.41	1.1048	1.1160	1.01

Table 3: Comparison of the relative output factor (ROF) for the 6-MV photon beams obtained from our MC simulation and from previous study[4].

Field size (cm^2)	MC simulation[4]	Our MC simulation	Percentage difference (%)
5×5	0.9020	0.9123	1.14
10×10	1.0000	1.0000	0.00
15×15	1.0610	1.0559	0.48
20×20	1.1050	1.0940	1.00
30×30	1.1590	1.1499	0.79

Table 4: Comparison the MC and TLD absolute doses for a clinical simple plan delivered on the CT-based data of the Rando phantom in the 6- and 10-MV photon beam.

No	Regions	6 MV				10 MV			
		MC dose (cGy)	TLD		Percentage difference (%)	MC dose (cGy)	TLD		Percentage difference (%)
			Dose (cGy)	STD (%)			Dose (cGy)	STD (%)	
AP-PA technique									
1	Lung (iso)	195.0	193.9	0.25	0.6	191.7	190.5	1.23	0.6
2	Lung	197.3	198.3	0.88	-0.5	194.5	193.9	1.23	0.3
3	Lung	195.3	196.4	0.94	-0.6	190.7	188.8	1.90	1.0
4	Tissue	203.9	203.5	0.95	0.2	198.0	196.3	0.72	0.9
5	Interface*	199.4	196.9	1.74	1.3	192.7	192.4	2.36	0.2
6	Interface*	189.7	186.8	0.68	1.5	186.2	184.8	0.76	0.8
7	Interface*	207.3	204.5	1.40	1.35	197.1	196.1	0.37	0.51
RAO-LPO technique									
1	Lung (iso)	192.2	190.5	1.10	0.9	188.9	187.2	0.68	0.9
2	Lung	195.0	193.0	2.03	1.0	190.2	187.1	2.03	1.7
3	Lung	184.8	183.9	1.19	0.5	181.1	180.0	2.66	0.6
4	Tissue	217.0	216.6	0.99	0.2	209.0	208.1	0.71	0.4
5	Interface*	195.8	194.7	0.71	0.6	191.8	190.5	0.56	0.7
6	Interface*	197.7	197.3	1.47	0.2	191.9	189.2	1.22	1.4
7	Interface*	187.9	186.2	1.81	0.9	183.8	181.6	1.05	1.2

*Interface between lung and tissue densities

Table 5: Comparison the MC and CCC absolute doses for a clinical simple plan delivered on the CT-based data of the Rando phantom in the 6- and 10-MV photon beam.

No	Regions	6 MV			10 MV		
		MC dose (cGy)	CCC dose (cGy)	Percentage difference (%)	MC dose (cGy)	CCC dose (cGy)	Percentage difference (%)
AP-PA technique							
1	Lung (iso)	195.0	200.4	-2.8	191.7	200.1	-4.4
2	Lung	197.3	203.3	-3.0	194.5	203.2	-4.5
3	Lung	195.3	201.2	-3.0	190.7	199.6	-4.7
4	Tissue	203.9	212.0	-4.0	198.0	205.8	-3.9
5	Interface*	199.4	207.6	-4.1	192.7	205.0	-6.4
6	Interface*	189.7	194.5	-2.5	186.2	196.0	-5.3
7	Interface*	207.3	214.1	-3.3	197.1	209.4	-6.2
RAO-LPO technique							
1	Lung (iso)	192.2	200.0	-4.1	188.9	199.9	-5.8
2	Lung	195.0	202.4	-3.8	190.2	201.4	-5.9
3	Lung	184.8	191.7	-3.7	181.1	191.5	-5.7
4	Tissue	217.0	221.6	-2.1	209.0	214.8	-2.8
5	Interface*	195.8	205.1	-4.8	191.8	203.2	-5.9
6	Interface*	197.7	204.8	-3.6	191.9	202.8	-5.7
7	Interface*	187.9	195.0	-3.8	183.8	194.4	-5.8

*Interface between lung and tissue densities

The absolute dose at the relevant depths in a water phantom can be computed using the percentage depth dose, dose profiles and relative output factors (ROF) accounting for the effect of field sizes. Commonly, the ROF is defined at a point (x, y, z) along the beam central axis in a water phantom. It is the ratio of the dose under the given field and the corresponding dose under a 10×10 cm^2 field for the same number of MU. In this study, the relative output factor measurements were taken in a water phantom at a depth of 10 cm for both 6-and 10-MV photon beams. The MC simulations were carried out with a statistical uncertainty of less than 1% in the voxel placed at the isocenter. The relative output factor test gave the result shown in Table 2.

We found that our MC calculation accurately produces measured relative output factors (ROF). These results are consistent with the measured data and the data from other investigators.[4,17,18] As shown in Table 3, our MC simulated ROFs for 6 MV photon beam energy agree with those of Popescu *et al.*[4], since they had performed reliable measurement at the same beam energy from similar medical linear accelerators (the Varian Clinac 21EX). Therefore, we conclude that our calculated ROFs using the MC simulation can be used for finding the absolute dose in the square fields with the side ranging from 5 to 30 cm.

3.2. Absolute dose comparison in CT-based data

The clinical conventional plans were calculated and measured in the chest region of the Rando phantom. To reduce the discrepancies of calculation in the CT-based data, the field size larger than 5×5 cm^2 were selected for both of 6- and 10-MV photon beams. The validation of the absorbed dose derived from the MC calculation in CT-based data was performed for a 10×12 cm^2 field for the AP-PA direction and an 8×15 cm^2 field for RAO-LPO direction. Table 4 compares the MC absolute dose calculated by equation (1) with the measured dose determined by a set of calibrated TLD placed inside the treatment field. The agreement of MC with TLD was evaluated using the point dose analysis. Our results showed that the MC absolute dose in the conventional treatment technique have acceptable agreement with the TLD in lung and interface region. The maximum difference was less than 2% for both of 6- and 10-MV photon beam.

In this work, the MC results were also compared to CCC doses in the same treatment plan. Table 5 lists the percent differences for the absolute doses comparison between the MC and the CCC data in the chest region of a Rando phantom for 6- and 10-MV photon beam energies. For 6-MV photon beam, the maximum difference for all treatment plans was 4.75%. This difference was 6.38% in the case of the 10-MV photon beam.

4. Discussion

As shown in the study of Lie *et al.*[6], the output of clinical accelerator is affected by the backscatter from jaws towards the monitor chamber. They found that the backscatter decreases approximately linearly with increasing field size. Therefore, an essential component of absolute dosimetry based on the MC simulation is the dose accumulated in the monitor chamber, which is used to control the radiation output for precise delivery of prescribed dose. Normally, the monitor chambers are calibrated in such a way that one monitor unit (1 MU) corresponds to 0.01 Gy at the depth of the maximum dose on the central axis in a water phantom under reference dosimetry conditions. Additionally, the correlation between the dose accumulated in the monitor chamber and the number of incident electron

on the linac target is also important. Therefore, in this study the forward and backscattered doses have been separated in the simulation in order to evaluate the accuracy of the derived absolute doses. As shown in Figures 1 and 2, the absolute dose for both simulated and measured data increased linearly with increasing field sizes, likely due to increasing amounts of electrons scattering from the collimator, air, and phantom. The higher difference can be found in the square field of 5×5 cm^2 at the depth of maximum dose because our matching condition between the simulated and measured data has been only performed for the 20x20 cm^2 field size. The calculated output for the 5×5 cm^2 field was overestimated in this depth. For the 6-MV photon beam, the measured dose at the field size of 5×5 cm^2 obtained from this dosimeter was 4% higher than that simulated dose. The discrepancy was reduced to 1.3% for the 10-MV photon beam. It was seen that this effect is more pronounced in low energy photon beam, because the effect of electron contamination in the shallow depth of maximum dose for the 6-MV, 1.5 cm depth, is higher than that for the 10-MV photon beam, 2.5 cm depth.

Comparing with the MC calculations, the CCC doses show a large overestimation in all studied points. We found that the discrepancies were distributed depending on the positions and energies. As seen in Table 5, the largest differences between the MC and the CCC data occur in the interface region for both of our photon beam energies. The overestimations for 10 MV were higher than 6 MV both in the lung and interface region. These results are similar to those found by other researchers.[1,19-21] While the MC doses near the interface region were comparable to those from TLD measurement, the CCC doses have notable differences, which is consistent with the reports of Han *et al*.[1] Because the presence of large heterogeneities is not accurately accounted for by the superposition-convolution algorithm, the dose calculation uncertainties of CCC mostly occur near the interface of materials with large density differences. The limitation of the CCC dose calculation in this region is due to its inability to model the backscattered photons and backscattered secondary electrons originating from both upstream and downstream tissues across the interface. In this study, we are aware of our CCC calculation limitations using the identical grid size to that of MC simulation. This may be considered too big in practical treatment planning.

5. Conclusion

For the depth beyond the depth of maximum dose, the MC-absolute dose for these 6- and 10-MV photon beams were in good agreement with the measurements from similar machines. Our calculated ROFs using the MC simulation were justified and can be used for finding the absolute dose in the square field with its side ranges from 5 to 30 cm. The absolute doses based on the MC data in conventional treatment technique have acceptable agreement with the TLD in lung and interface region of CT Rando phantom set. The maximum difference was less than 2% for both of 6- and 10-MV photon beam. In contrary, the large variability was also observed at the interface region between the MC-absolute doses and the CCC dose calculations.

Conflict of Interest

The authors declare that they have no conflicts of interest. The authors alone are responsible for the content and writing of the paper.

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