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Dose calculation accuracy for photon small fields in treatment planning systems with comparison by Monte Carlo simulations

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Abstract

Purpose: Advanced radiation therapy techniques use small fields in treatment planning and delivery. Small fields have the advantage of more accurate dose delivery, but with the cost of some complications in dosimetry. Different dose calculation algorithms imported in various treatment planning systems (TPSs) which each of them has different accuracy. Monte Carlo (MC) simulation has been reported as one of the accurate methods for calculating dose distribution in radiation therapy. The aim of this study was the evaluation of TPS dose calculation algorithms in small fields against 2 MC codes.

Methods: A linac head was simulated in 2 MC codes, MCNPX, and GATE. Then three small fields (0.5×0.5, 1×1 and 1.5×1.5 cm²) were simulated with 2 MC codes, and also these fields were planned with different dose calculation algorithms in Isogray and Monaco TPS. PDDs and lateral dose profiles were extracted and compared between MC simulations and dose calculation algorithms.

Results: For 0.5×0.5 cm² field mean differences in PDDs with MCNPX were 2.28, 4.6, 5.3, and 7.4% and with GATE were -0.29, 2.3, 3 and 5% for CCC, superposition, FFT and Clarkson algorithms respectively. For 1×1 cm² field mean differences in PDDs with MCNPX were 1.58, 0.6, 1.1 and 1.4% and with GATE were 0.77, 0.1, 0.6 and 0.9% for CCC, superposition, FFT and Clarkson algorithms respectively. For 1.5×1.5 cm² field mean differences in PDDs with MCNPX were 0.82, 0.4, 0.6 and -0.4% and with GATE were 2.38, 2.5, 2.7 and 1.7% for CCC, superposition, FFT and Clarkson algorithms respectively.

Conclusions: Different dose calculation algorithms were evaluated and compared with MC simulation in small fields. Mean differences with MC simulation decreased with the increase of field sizes for all algorithms.

Key words: small field; radiation therapy; Monte Carlo simulation; MCNPX; GATE; dose calculation algorithms.

Introduction

Advanced radiation therapy techniques such as IMRT, SBRT, and VMAT are using small photon beamlets (smaller than 4×4 cm²) for treatment planning and treatment delivery.¹⁻³ Steep dose gradient in these fields results in a higher dose to target volume and simultaneously less dose to organs at risk (OARs),⁴⁻⁶ this is the ultimate goal in radiation therapy.⁵ However, using small fields will make some trouble in beam dosimetry.¹ The partial occlusion of primary photon beam from source, loss of lateral charged particle equilibrium, steep dose gradient and volume averaging in dosimeter are the major issues that make dosimetry in the small field a challenging work.^{1,2,7-9} Because of these problems, there is no ideal dosimeter for use in small fields.⁹ Tissue equivalent, energy and dose rate independence, linear response, high resolution and small relative size are the ideal characteristics of a suitable detector for the dosimetry of small fields.¹ Various dosimeters have been used for small field

dosimetry despite all challenges and limitations, such as small volume ion chambers, radiochromic films, diodes, diamond detectors, and plastic scintillators.^{1,10,11}

Monte Carlo (MC) simulation was used to calculate dose distribution in small field radiotherapy, in various studies.^{2,12-14} Since MC simulation takes the exact geometry of linac, electron transport, and beam configurations into account,¹⁵⁻¹⁶ its precision does not limit by complications from advanced techniques (such as IMRT and small fields) and tissue heterogeneities.¹⁵ It was mentioned that MC simulation is one of the most accurate approaches to calculate dose distribution in radiation therapy dosimetry.^{1,2,12,15-17} MC techniques are accepted as the gold standard in radiotherapy dose calculation.^{1,18} It is known that MC will give accurate dose calculation in regions of charged particle disequilibrium and where measurement interpretation is challenging.¹ So MC simulation has considered as standard practice in benchmarking dose calculation algorithms used in

commercial treatment planning systems.^{1,6,12} However MC simulation accuracy is limited by time of calculation,¹⁹ and this is why these algorithms aren't used widely in commercial treatment planning systems.²⁰ There are several different MC codes available for simulating a radiation therapy unit, including EGSnrc, ETRAN, Geant4, and MCNPX.²¹ MCNPX is widely used to simulate medical linear accelerators employed in radiotherapy.^{2,22} GATE (Geant4 Application for Emission Tomography) is a Monte Carlo platform that helped neuromas studies in nuclear medicine. A recent version of this code enabled the modeling of X-ray computed tomography and radiotherapy.²³ Sarrut et al demonstrated the potential and versatility of this code for dosimetry and radiotherapy applications since it's open-source and user-friendly.²⁴ It is expected that GATE will have an increasing role in combined therapy and imaging.

The maximum error in dose delivery in the entire radiotherapy process is desired to be less than 5%,^{16,18,25,26} therefore it needs all steps of this process to be in an optimal situation to decrease the error as much as possible. Treatment planning is an important step in this process and the dose calculation algorithm is the most important part of it, which determines the accuracy of the treatment planning system. There are two types of dose calculation algorithms: measurement-based (such as pencil beam convolution (PBC) algorithm) and model-based (such as anisotropic analytical algorithm (AAA), collapsed cone convolution (CCC) and Acuros XB (AXB) algorithm).^{1,6} Small field dosimetry issues are the origin of inaccuracies in dose calculation algorithms since measurement data is required in algorithm configuration.^{6,27} This problem has notable outcomes in the accuracy of measurement-based algorithms. But in model-based algorithms inherent performance of the algorithm determines the accuracy of dose calculation.⁶ These inaccuracies are mostly due to the inability of the algorithm to model the loss of electronic equilibrium,^{6,15,16} which are specific to measurement-based algorithms that are usually based on pencil beam convolution.²⁸ More advanced algorithms, model-based, which are based on superposition convolution principles, have more acceptable results since they model the loss of charged particle disequilibrium.²⁸ It was shown that PBC algorithm (measurement-based) overestimate dose to lung inhomogeneities and small fields (up to 30% in some cases) where the loss of lateral charged particle equilibrium happens.^{16,18,29,30} However model-based algorithms such as AAA and CCC have reliable and more accurate dose calculations.^{1,15,27,30,31} There is an Acuros XB algorithm that solves the linear Boltzmann transport equation which describes the macroscopic conduct of radiation particles throughout the matter.³² Several studies have compared the AAA algorithm with AXB and MC simulation.^{1,32} Results showed that AXB has superior accuracy in comparison with AAA in the region of electronic disequilibrium and tissue heterogeneities.³¹ These differences are reported to be due to beam energy, field sizes and density of materials.³³ Fogliata and Cozzi made a review of

different types of dose calculation algorithms that are being used in SBRT TPS for lung cancer cases.²⁸ They concluded that measurement-based algorithms aren't suitable for such cases (small fields and low density) which they show differences in the order of 20-30%, while model-based and more advanced (AXB) algorithms provide more accurate results and have differences in the order of 10% and 5% respectively.

The aim of the current study was to evaluate the performance of four dose calculation algorithms: collapsed cone convolution (CCC) employed in Monaco treatment planning system, superposition-convolution, fast Fourier transform (FFT) and Clarkson used in Isogray TPS in comparison with Monte Carlo simulation in small photon fields in a homogeneous water phantom. We employed 2 MC codes, MCNPX and GATE.

Materials and methods

MC simulation

MC simulations are a type of mathematical method for calculating dose distribution by transporting electrons and photons based on principles of radiation transport. In this study, we used two different MC codes. Models for both codes are described below.

A. MCNPX

MCNPX Monte Carlo code³⁴ was used to simulate Elekta Synergy Platform linear accelerator. Linacs have a modular construction that is specified by different vendors. The vendor's specified data was used to simulate all necessary parts of linac's head, including electron target, primary collimator, flattening filter, ionization chambers, mirror, and secondary definition jaws. The model also consisted of a 40×40×40 cm³ cylindrical water phantom. 6 MeV photon beam was used for simulation. This model was divided into two parts: 1) patient independent parts (including electron target, primary collimator, flattening filter, ionization chambers and mirror), and 2) patient dependent part (including secondary collimators and water phantom). By scoring photons that cross a plane above secondary collimators a phase space file was generated (represent patient independent parts). Then the phase space was used as a photon point source at 100 cm source to surface distance (SSD). For different calculation purposes, different field sizes were required which were designed by changing the opening of secondary collimators. Initial electron energy was benchmarked by comparing the measured and calculated data.

For model verification, we simulated 3 larger fields (6×6, 10×10 and 20×20 cm²), where challenges due to small field complexity do not exist.² For these fields, PDDs and beam profiles were calculated and compared with direct measurement data. Direct measurements were done by a 0.125 cm³ Semiflex ion chamber (type 31010, PTW-Freiburg, Germany) in a MP3 water phantom (PTW-Freiburg, Germany).

For comparison with treatment planning system outputs, three small fields (0.5×0.5 , 1×1 and 1.5×1.5 cm²) were simulated. To calculate PDDs in the water phantom, a set of cylindrical scoring cells with a radius equal to 10% of the beam's diameter³⁵ and 2 mm height was used along the beam's central axis. For profile calculation, a set of scoring cells with $2 \times 3 \times 3$ mm³ dimensions was used in the depth of maximum dose and 10 cm across the central axis. By running particle histories through the water phantom, deposited energy in scoring cells was calculated by *F8 tally. Electron and photon energy cut-off was set to 0.4 and 0.01 MeV respectively. Calculations were done on a workstation, by running about 2×10^9 histories for each field size, the statistical uncertainty was less than 0.5% for all of the calculations.

B. GATE simulation

GATE is an advanced open-source software package developed by OpenGATE collaboration dedicated to numerical simulation in medical imaging and radiation therapy. This code is user-friendly and uses command scripts. GATE is based on Geant4 and uses its libraries. The radiation therapy option is added to this code in version 6.2. This code considers all physical phenomena and the user can determine which physical phenomenon to be considered in the physics list input file. Emstandard_OPT3 physics list from Geant4 was used in our model. In this study, version 8.2 was used to simulate Elekta Synergy Platform linear accelerator. Geometry data was provided by the vendor. All necessary parts were simulated. The model included an electron target, primary collimator, flattening filter, ionization chambers, mirror, secondary collimators, and a $40 \times 40 \times 40$ cm³ homogeneous water phantom. Different field sizes were simulated for further calculation. The initial electron energy was 6 MeV. For energy benchmarking, direct measurement data was used, this data was the same as those were used for MCNPX benchmarking. Model verification was carried out similarly to the MCNPX model.

In GATE just like simulation in MCNPX, simulation was divided into two parts, patient independent part and patient dependent part. Like MCNPX a phase space file was generated by scoring the photons that crossing a plane above secondary collimators. Then the phase space file was used as a point source at 100 cm source to surface distance. Same as MCNPX, three small fields (0.5×0.5 , 1×1 and 1.5×1.5 cm²) were simulated. Cut-offs were set to 2 mm for photons and 1 mm for electrons, which are equal to 10 and 350 keV respectively. Calculating dose distribution was done by attaching an actor to the water phantom. This actor is Dose Actor. The Dose Actor output file is a 3D matrix in mhd or raw formats. The voxel size on this matrix was $2 \times 2 \times 2$ mm³. This actor has different output data, including dose, dose uncertainty, energy deposited, energy deposited uncertainty, etc. Users can define which one of these output data's to be active or not. For this work dose and dose uncertainty was needed. Then dose distribution file was imported into MATLAB software and PDDs and profiles were

extracted. After running about 2×10^9 particle histories, uncertainty in all fields was less than 2%.

TPS data acquisition

A. Dose calculation in Isogray TPS

Isogray version 4.1.3 treatment planning system (Dosisoft Co., France) was used in this study. This TPS has different dose calculation algorithms including superposition convolution, fast Fourier transform, and Clarkson. This TPS works with various techniques such as IMRT. Algorithm commissioning was carried out precisely using the vendor's instructions and recommendations. For this study superposition convolution, fast Fourier transform and Clarkson algorithms were used. The First 2 of these algorithms are model-based and the latest is measurement-based. Beam configurations were done for three small fields, 0.5×0.5 , 1×1 and 1.5×1.5 cm², where source to surface distance was 100 cm. dose distribution calculation was done in a $40 \times 40 \times 40$ cm³ homogeneous water phantom created in the treatment planning system. For all plans, 6 MeV beam energy was used. Dose calculation grid size was set to the minimum size available, which was $2 \times 2 \times 2$ mm³. All plans calculated dose distribution using all algorithms. To investigate calculated dose distribution, PDDs and profiles were extracted. PDDs extracted along the beam's central axis and profiles in 2 depths, depth of maximum dose, and 10 cm.

B. Dose calculation in Monaco TPS

Collapsed cone convolution (CCC) dose calculation algorithm implemented in Monaco 5.11.02 was also evaluated in the current study. This algorithm was commissioned by the manufactory's recommendation. All previous plans for different field sizes from Isogray were repeated in Monaco with the CCC dose calculation algorithm and with the dose calculation grid size of $2 \times 2 \times 2$ mm³.

Results

Validation of Monte Carlo simulation

A new formalism and machine-specific reference field for a situation that cannot establish conventionally were introduced by Alfonso et al. for calibration intentions.³ In this study, a standard reference field 10×10 , also 20×20 and 6×6 cm² were used as machine-specific reference fields for model validation. 6×6 cm² field was used because its closer to small fields and charged particle equilibrium still exists. Primary electron source energy was determined by trial and error by comparing PDD and Profiles (in the depth of 10 cm) from simulation and measurement in a homogeneous water phantom and 10×10 cm² field size. **Figure 1** shows a comparison in PDDs between 2 MC simulations (MCNPX and GATE) and measurement for 6×6 , 10×10 , and 20×20 cm² fields. For the 6×6 cm² field, the mean differences between MCNPX and GATE with measurement were 0.59% and -0.2% respectively with the maximum difference (-14% and 7% respectively) being in the buildup

region. For the $10 \times 10 \text{ cm}^2$ field, the mean differences were -0.39% and -0.5% respectively and for the $20 \times 20 \text{ cm}^2$ field, the mean differences were -2.4% and -1.9% respectively. **Figure 2** shows beam profiles for 20×20 , 10×10 , and $6 \times 6 \text{ cm}^2$ Fields. Statistical uncertainty for both MC codes was below 0.5%. Venselaar et al. divided dose profiles into three regions: 1) the umbra or flat region inside the field, 2) the penumbra region and 3) the low dose region outside the field.³⁶ There are several recommendations for evaluating calculations in different situations, one of them introduced by Venselaar et al. where recommend criteria for acceptance of calculation as follow, for 1) umbra region 2% difference, 2) for penumbra region 10% and for 3) low dose region 30%.³⁶ For the $6 \times 6 \text{ cm}^2$ field, there were the mean difference of -0.36% and 0.009% inside the field for MCNPX and GATE respectively. For the penumbra region, these differences were -9.6% for MCNPX and -10.4% for GATE. And finally, for the low dose region mean differences

were -28.8% and -18.2% for MCNPX and GATE. For $10 \times 10 \text{ cm}^2$ field, the mean differences were -1% and -0.6% for flat region, -2.5% and -9.8% for penumbra region and -28.7% and -19.5% for low dose region for MCNPX and GATE respectively and for $20 \times 20 \text{ cm}^2$ field, the mean differences were -0.3% and -0.4% for flat region, -17.6% and -17.7% for penumbra region and -28.4% and -28.7% for low dose region for MCNPX and GATE respectively. For the $6 \times 6 \text{ cm}^2$ field, full width at half maximum (FWHM) was 6.75, 6.68 and 6.62 cm for measurement, MCNPX and GATE respectively. For the $10 \times 10 \text{ cm}^2$ field, FWHM was 11.06, 11.09, and 11 cm for measurement, MCNPX, and GATE respectively. For the $20 \times 20 \text{ cm}^2$ field, FWHM was 23.89, 23.91, and 23.92 cm for measurement, MCNPX, and GATE respectively. Differences in FWHM were less than 2 mm between measurement and calculations for all fields.

Figure 1. PDD curves in a homogenous water phantom, a comparison between measurements, MCNPX and GATE for various field sizes. A) $6 \times 6 \text{ cm}^2$ and B) $10 \times 10 \text{ cm}^2$. C) $20 \times 20 \text{ cm}^2$.

Figure 2. Lateral beam profiles in a homogenous water phantom, a comparison between measurements, MCNPX and GATE for various field sizes. A) $6 \times 6 \text{ cm}^2$ and B) $10 \times 10 \text{ cm}^2$. C) $20 \times 20 \text{ cm}^2$.

Comparing PDDs between TPSs and MCNPX and Gate in a small field

In this study, three small fields were investigated between 2 TPSs and Monte Carlo simulations, 0.5×0.5 , 1×1 , and 1.5×1.5 cm². Calculations for both Monte Carlo codes were done for 2×10^9 histories and statistical uncertainty was below 1% for all situations and fields. **Figure 3** shows a comparison between PDDs for dose calculation algorithms and MCNPX and GATE for different field sizes. Four dose calculation algorithms were evaluated in the current study, CCC, superposition, FFT and Clarkson. The First three of these algorithms are model-based and the latest is correction-based. For 0.5×0.5 cm² field mean

differences in PDDs with MCNPX were 2.28%, 4.6%, 5.3% and 7.4% and with GATE were -0.29%, 2.3%, 3% and 5% for CCC, superposition, FFT and Clarkson algorithms respectively. For 1×1 cm² field, the mean differences in PDDs with MCNPX were 1.58%, 0.6%, 1.1% and 1.4% and with GATE were 0.77%, 0.1%, 0.6% and 0.9% for CCC, superposition, FFT and Clarkson algorithms respectively. For 1.5×1.5 cm² field, the mean differences in PDDs with MCNPX were 0.82%, 0.4%, 0.6% and -0.4% and with GATE were 2.38%, 2.5%, 2.7% and 1.7% for CCC, superposition, FFT and Clarkson algorithms respectively. Relative depth doses for 0.5×0.5 , 1×1 , 1.5×1.5 cm² field are shown in **Tables 1, 2** and **3** respectively.

Figure 3. PDD curves for small fields in a homogenous water phantom, a comparison between MCNPX, GATE and various dose calculation algorithms in different field sizes. A) 0.5×0.5 cm², B) 1×1 cm² and C) 1.5×1.5 cm².

Table 1. Relative depth doses (percentage) calculated by different dose calculation algorithms and 2 Monte Carlo codes (MCNPX and GATE) for 0.5×0.5 cm² field size.

Depth (cm)	MCNPX (%)	GATE (%)	CCC (%)	Superposition (%)	FFT (%)	Clarkson (%)
0	46.36	45.43	32.14	32.4	32	30.37
1.1	100	100	99.68	97.74	98.67	97.29
5	77.69	80.23	78.99	77.3	78.27	83.31
10	55.92	57.58	57.41	57.37	57.87	60.92
15	40.69	41.79	42.61	42.76	43.07	44.64
20	29.94	30.69	31.36	32.14	32.27	32.97

Table 2. Relative depth doses (percentage) calculated by different dose calculation algorithms and 2 Monte Carlo codes (MCNPX and GATE) for 1×1 cm² field size.

Depth (cm)	MCNPX (%)	GATE (%)	CCC (%)	Superposition (%)	FFT (%)	Clarkson (%)
0	39.61	40.59	53.52	31.81	31.51	31.28
1.3	100	100	100	97.54	98.55	99.57
5	79.61	80.75	79.07	77.79	78.62	82.47
10	57.61	57.83	57.91	57.14	57.57	59.62
15	41.98	42.29	43.15	42.19	42.43	43.15
20	30.92	30.94	31.8	31.47	31.51	31.56

Table 3. Relative depth doses (percentage) calculated by different dose calculation algorithms and 2 Monte Carlo codes (MCNPX and GATE) for 1.5×1.5 cm² field size.

Depth (cm)	MCNPX (%)	GATE (%)	CCC (%)	Superposition (%)	FFT (%)	Clarkson (%)
0	37.89	37.69	51.99	29.36	29.44	32.24
1.5	100	100	99.05	98.87	99.28	99.8
5	80.74	80.15	79.77	79.98	80.31	82.25
10	58.5	57.68	58.49	58.32	58.56	59.33
15	42.67	41.62	43.24	42.81	42.87	42.9
20	31.39	30.61	31.84	31.62	31.69	31.34

Comparing dose profiles between TPS and MCNPX and Gate in the small field

For comparing dose profiles in small fields, profiles in 2 depths were extracted: depth of $D_{\max}$ and 10 cm in a homogeneous water phantom. Profiles at $D_{\max}$ are presented in **Figure 4**. For 0.5×0.5 cm² field, FWHMs were calculated 0.5266, 0.5161, 0.9672, 0.7385, 0.7136 and 0.5749 cm by MCNPX, GATE, CCC, superposition, FFT and Clarkson respectively. All discussed dose calculation algorithms overestimate FWHM against MCNPX or GATE. Differences were up to 0.4511 cm. Errors in calculating FWHM against MCNPX were 83.6, 40.2, 35.5, and 9.2% and against GATE were 87.4, 43.1, 38.3, and 11.4 for CC, superposition, FFT and Clarkson respectively. For 1×1 cm² field, FWHMs were calculated 1.0581, 1.0438, 1.1548, 1.182, 1.1673 and 0.971 cm by MCNPX, GATE, CCC, superposition, FFT and Clarkson respectively. In this field, differences were up

to 0.1382 cm. Differences with MCNPX were 9.4% 11.7%, 10.3% and -8.2% and with GATE were 10.6%, 13.2%, 11.8%, and -7% for CCC, superposition, FFT and Clarkson respectively. For the 1.5×1.5 cm² field, FWHM was calculated 1.5598, 1.5532, 1.4431, 1.576, 1.5686 and 1.4198 cm by MCNPX, GATE, CCC, superposition, FFT and Clarkson respectively. In this field, differences were up to -0.1334 cm which is for the Clarkson algorithm. Differences in FWHM with MCNPX were -7.5%, 1%, 0.6% and -9% and with GATE were -7.1%, 1.5%, 1% and -8.6% for CCC, superposition, FFT and Clarkson respectively. Profiles for depth 10 cm through homogeneous water phantom are presented in **Figure 5**. In this depth, there were some differences in FWHM calculated by dose calculation algorithms and Monte Carlo simulation. Calculated FWHM for different algorithms and field sizes are presented in **Table 4**.

Figure 4. Lateral dose profiles for small fields in a homogenous water phantom at depth 1.5 cm across beam's central axis, a comparison between MCNPX, GATE and various dose calculation algorithms in different field sizes. A) 0.5×0.5 cm², B) 1×1 cm² and C) 1.5×1.5 cm².

Figures5. Lateral dose profiles for small fields in a homogenous water phantom at depth 10 cm across beam's central axis, a comparison between MCNPX, GATE and various dose calculation algorithms in different field sizes. A) 0.5×0.5 cm², B) 1×1 cm² and C) 1.5×1.5 cm².

Table 4. FWHM values (in cm) for different dose calculation algorithms in various field sizes and depths.

Field size	Depth (cm)	MCNPX (cm)	GATE (cm)	CCC (cm)	Superposition (cm)	FFT (cm)	Clarkson (cm)
0.5×0.5 cm ²	1.5	0.527	0.516	0.967	0.738	0.713	0.575
	10	0.5848	0.568	1.026	0.81	0.7801	0.6147
1×1 cm ²	1.5	1.0581	1.043	1.154	1.182	1.167	0.971
	10	1.153	1.1415	1.326	1.2845	1.2677	1.0782
1.5×1.5 cm ²	1.5	1.5598	1.5532	1.4431	1.576	1.5686	1.4198
	10	1.66	1.6472	1.643	1.7159	1.7079	1.5563

Discussion

The use of modern radiation therapy techniques such as IMRT, VMAT, SRS, SBRT, and IGRT had led to superior accuracy in dose delivery. These techniques use small beamlets in treatment delivery and employ complex dose calculation algorithms in treatment planning systems. In IMRT and VMAT, intensity modulation is achieved using small fields produced by multileaf collimators (MLC) routinely. In the VMAT technique, gantry rotation and dose rate also are free degrees and its complexity is more than IMRT. So the requirement for dose verification of small fields is more important.

Small fields have a high dose gradient that delivers a high dose to the tumor and spare surrounding normal tissues. So determining dose in small beams is highly desired, but because of loss of lateral charged particle equilibrium, partial occlusion of primary photons from source and size of detectors, dosimetry in the small field is difficult. Therefore, it is important to investigate their effect on dose calculation algorithms used in treatment planning systems. Various publications have discussed and investigated the accuracy of different dose calculation algorithms against MC simulations or direct measurement with different dosimeters for small field sizes.^{1,2,6,18,27,28,31-33,37-39} The goal of this article was to investigate the accuracy of CCC, superposition, FFT and Clarkson dose calculation algorithms in small photon fields against two Monte Carlo codes: MCNPX and GATE. Validations of two models were carried out by using 3 fields (6×6, 10×10 and 20×20 cm²). PDD results showed a good agreement between MC simulations and measurements by the ionization chamber (under 2.5% difference for all fields and codes). For the buildup region, a local differences of up to 15% for all fields and codes were observed. These differences also reported in other publications² and reported to be because of reading uncertainties in the ionization chamber.^{2,22} Also it might be for volume averaging in dosimeter because of the high dose gradient in this region.¹⁷ For lateral dose profiles, our results for 6×6, 10×10 and 20×20 cm² fields meet with the criteria purposed by Venselaar et al.³⁶ and Mesbahi et al.²² for MC codes very well.

As mentioned there are 2 types of dose calculation algorithms that are used in the treatment planning systems. Between the four algorithms used in this study, Clarkson is measurement-based, CCC, superposition, and FFT are model-based. We investigate the accuracy of these algorithms in a homogeneous water phantom against MC simulation. For Clarkson algorithm

differences up to 7.5% were observed in PDDs, smaller discrepancies were observed for larger fields. CC algorithm overestimated dose in PDDs against MC simulations for all fields up to 2.5% which decreased with the increase of field sizes. Superposition and FFT algorithm showed similar behaviors, they both overestimated dose in PDD curves which differences are maximum for smaller field size. Similar results were reported in various publications for different types of algorithms. The FWHMs obtained from discussed algorithms had more differences in smaller fields, and CCC had the most difference in 0.5×0.5 cm². These differences showed the inability of these algorithms to calculate dose profiles in so small fields. When the field size increased, the differences in the FWHM decreased dramatically. This is in an agreement with previous studies.¹ Behinaein et al investigated the accuracy of AAA and AXB algorithms against measurement with plastic scintillator and Monte Carlo simulation both inhomogeneous and heterogeneous situations for small photon fields.¹ They used PDD and Profiles to compare algorithms. In homogeneous water phantom, they observed both algorithms agreed well with MC simulation and measurement with the plastic scintillator. Also, they reported that in a heterogeneous situation both algorithms have a good dose prediction comparing to MC simulation and measurement, although they mentioned a superior accuracy for the AXB algorithm. Chopra et al evaluated the accuracy of 5 dose calculation algorithms used in 3 treatment planning systems in the presence of tissue heterogeneous for a range of field sizes, from small to large.⁴⁰ They used the ion chamber and EDR film for PDD and dose profile measurements in a range of field sizes. They observed that an MC algorithm used in Brainlab TPS has a good agreement compared to measurements (less than 3%), and agreement improved with the increase of field size. Although the PB algorithm in this TPS significantly over predicted dose against the measurement (up to 34%). Their results showed differences between PB and measurements which increased with the decrease of field size, decreasing phantom density and increasing depth within heterogeneity. AAA algorithm used in Eclipse TPS, over predicted dose compare to measurements, on the other hand, the AXB algorithm showed a good agreement with measurement for heterogeneous media as well as for interface region. Ojala et al evaluated the accuracy of 3 dose calculation algorithms, PBC, AAA, and AXB which were used in the Eclipse treatment planning system for several small tumors placed in lung patients.⁶ They used dose distribution from a full Monte Carlo

simulation as a reference. AXB showed a good agreement with MC simulation but for smaller PTVs, larger discrepancies observed. For AAA and PBC mean dose to smaller sized PTVs had discrepancies up to 20% and 60% respectively. Stathakis et al investigated the accuracy of the AXB algorithm against AAA, CCC, and MC simulation in a homogeneous and heterogeneous medium for a set of small fields.⁴¹ Their results showed good agreement for all algorithms in homogeneous phantom which differences were $\pm 1.5\%$. For Heterogeneous phantom, AXB and CCC showed a good agreement with MC simulation (less than 2% difference) where for AAA differences were up to 5%. Calvo et al evaluated the CCC algorithm (Pinnacle) for lesions with a diameter smaller than 3 cm.³⁷ Their results demonstrated a 5.6% difference in PTV dose between CCC and MC simulation. Fotina et al conducted a comparison between CCC (Master Plan) and MC simulation for SBRT and IMRT cases.³⁹ The difference in PTV dose was reported 5% among the CCC dose calculation algorithm and MC simulation.

As declared our results for measurement-based and model-based algorithms were in an agreement with previous studies. For the Clarkson algorithm, our results showed a much smaller deviation with MC simulation than previous publications,^{6,40} which is because of this fact that those studies evaluated a measurement-based algorithm in a heterogeneous situation (mostly lung case study) but in this study, such an algorithm was evaluated in a homogeneous water phantom. Model-based algorithms in this study showed an acceptable agreement with MC simulation (less than 5% difference). CCC algorithm had a very large difference in the buildup region which is due to the fact that this algorithm does not model the buildup region when entering from a low density to a high-density region.⁴² All evaluated algorithms had better results in bigger fields and with a decrease of field size, deviations with MC simulation increased, as reported in previous studies.^{6,40,42} The source of these differences in small fields is reported to be the loss of lateral charged particle equilibrium.^{42,43} As reported by different authors most deviations appear in fields smaller than $1 \times 1 \text{ cm}^2$. Palmans et al have stated that partial occlusion of primary photons from source happens when the field size is comparable with focal spot size, also they mentioned that in medical accelerators focal spot size is about 0.5 cm.⁴⁴ So in fields smaller than $1 \times 1 \text{ cm}^2$, when the field size is comparable with focal spot size, partial occlusion of primary photons appears. AXB

algorithm is reported to have superior accuracy against other model-based algorithms (like AAA and CCC) in various publications.^{1,6,40,41} Model-based algorithms use precalculated dose kernels, while the AXB algorithm has a more detailed model and like MC uses direct models and solves physical interaction between particles. This algorithm accurately models primary photons from source and also scattered photon and electron fluence, for this job it solves the linear Boltzmann transform equation.⁴²

All model-based algorithms that investigated had an agreement with MC simulation in small photon fields in a homogenous media concerning this that CCC algorithm from Monaco TPS had a superior accuracy toward the others. Clarkson algorithm as we discussed as a correction based algorithm had good accuracy in larger fields and homogenous media, but with decrease of field size and appearing of loss of charged particle equilibrium its accuracy drops dramatically. For better evaluation of these algorithms for small fields, they must be challenged in a more complex situation like a heterogeneous media.

Conclusion

Different dosed calculation algorithms were evaluated and compared with MC simulation in small fields. Comparison is done with dosimetric parameters such as PDDs and dose profiles in a homogenous water phantom. Clarkson algorithm (measurement-based) showed a large discrepancy with MC simulation (up to 7.5%). Model-based algorithms had better agreement with MC simulations. CCC dose calculation algorithm used in Monaco showed superior accuracy than superposition and FFT employed in Isogray TPS. For all algorithms, mean differences decreased with the increase of field size which is in an agreement with previous publications.

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