Original Article

Determination of Dose Distributions by Monte-Carlo Simulation of 6 MV Photon Beam of Varian VitalBeam Accelerator Using Geant4 Multithreaded Code

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Abstract

Background: Accuracy of dose delivery in radiation therapy is a primary requirement for effective cancer treatment. In practice, dose delivery accuracy of ±5% is desired. To achieve this accuracy level, an accurate method for calculating the dose distributions in the tumor volume is required. Monte-Carlo method is one of the methods considered to be the most accurate for calculating dose distributions. **Materials and Methods:** G4 linac-MT code was used to simulate a 6 MV photon beam. The initial electron beam parameters were tuned to validate the beam modeling from depth doses and beam profile. The dose distributions measured in water phantom were compared to the calculated dose distributions based on gamma index criterion. **Results:** The beam tuning showed the initial electron energy, sigma and full width at half maximum of 6.2 MeV, 0.8 MeV, and 1.18 mm, respectively, best match the measured dose distributions. The gamma index tests showed the calculated depth doses and beam profile were generally comparable with measurements, passing the standard acceptance criterion of 2%/2 mm. The simulated photon beam was justified by the index of beam quality, which showed excellent agreement with measured doses with a discrepancy of 0.1%. **Conclusion:** The observed agreement confirm the accuracy of the simulated 6 MV photon beam. It can therefore be used as radiation source for calculating dose distributions and further investigations aimed at improving dose delivery and planning in cancer patients.

Keywords: Beam profile, Monte-Carlo simulation and G4 linac-MT, percentage depth dose, vital beam

Received on: 16-11-2021 **Review completed on: 03-02-2022** **Accepted on: 22-02-2022** **Published on: 05-08-2022**

Introduction

External-beam radiation therapy plays a key role in the curative intervention of cancer. The effectiveness of radiation therapy depends on the delivery of the highest possible damage to the tumor volume while sparing the normal tissues. Among other factors, accurate dose in amount and uniform distributions within the tumor volume during the course of radiation therapy is desired for curative intervention. In practice, an accuracy of ± 5% for the delivery of the prescribed dose is required.[1] To achieve this level of accuracy, a reliable method for dose calculation of better than ±2% is desired. Among many dose calculation algorithms implemented in treatment planning systems, Monte-Carlo (MC) method has been proven to be the most accurate technique for dose calculation in radiation therapy.[2,3] Thus, the use of MC simulation has been of great interest for accurate dosimetry. However, the major limitation to clinical implementation of MC to the routine practice is long simulation time for the dose calculation.[3] Luckily, significant improvement in processors and availability of computer architecture has tremendously reduced the simulation time.

In practice, the primary objective of MC simulation in dosimetry is to generate information on distribution and amount of dose in a defined geometry. Often, it is used when measurements are not feasible or reaches the limitation. To achieve this objective, one requires information of parameters related to radiation source and the object in which photon interactions producing the dose are taking place. The parameters related to the source create a beam model, which provides the Phase Space (Phsp) data. The Phsp data are then used as source input in the simulation code for performing

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DOI: 10.4103/jmp.jmp_139_21

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How to cite this article: Chiuyo JS, Lugendo IJ, Muhogora WE. Determination of dose distributions by monte-carlo simulation of 6 MV photon beam of varian vitalbeam accelerator using geant4 multithreaded code. J Med Phys 2022;47:181-8.

dose calculation in a particular geometry. In order to achieve the desired accuracy of photon beam simulation and consequently dose calculation requires precise specifications of the accelerator head geometry and parameterization of the initial electron beam that produces bremsstrahlung photons. However, for proprietary reasons, such specifications are rarely provided by the manufacturers.[4] This makes the geometrical specifications unavailable to most researchers. This is the case of Varian VitalBeam linear accelerator (linac) at Ocean Road Cancer Institute (ORCI) in Tanzania. Although the Phsp can be made available for MC simulation, among other limitation of these Phsp files is that the information of the initial electron beam parameters are unknown.[4] These initial electron beam parameters cannot directly be adopted to match dose measured in the user's linac, because they vary even among linacs of the same model.

To the best of our knowledge, few studies have been conducted on the validation of Geant4 multithreaded code on different linac like TrueBeam/VitalBeam Varian medical systems.[2] The objective of this study therefore was to determine the dose distributions by simulating a 6 MV photon beam delivered by Varian VitalBeam linac using multithreaded G4 linac-MT platform. To achieve this objective, the calculated dose distributions are validated with measurements. The results obtained are of great importance for further investigations aimed at improving dose delivery and planning in cancer patients at ORCI.

Materials and Methods

Reference data for dose comparison

As an indispensable requirement, simulation must be experimentally benchmarked against measured data. These requirements are often achieved using phantom measurements with simple geometries. The measurement of depth doses and beam profiles often in a water box geometry phantom is among the most widely used parameters in a variety of photon beams. To achieve this requirement, the measurement of depth doses and dose profile was performed using 6 MV photon beam of a Varian VitalBeam linac (Varian Medical systems, Inc, Palo Alto, CA) installed at ORCI. The acquisition of beam data was done based on Technical Report Series (TRS)-398 recommendations.[5] The depth doses and beam profile were determined using 10 cm × 10 cm in a water phantom of 40 cm × 40 cm × 40 cm at 100 cm source to surface distance (SSD). The doses were measured using 0.125 cc semiflex ionization chamber (PTW-31010, Freiburg, Germany) connected to a UNIDOS electrometer calibrated at International Atomic Energy Agency (IAEA) dosimetry laboratory. The calibration of the ionization chamber is traceable to Physikalisch-Technische Bundesan standards. The depth doses were measured along the central axis (CAX) of the photon beam at depths ranging from 0.5 to 30 cm, with a step of 0.5 cm. The cross-line beam profile was measured at 10 cm depth by moving the chamber horizontally across the CAX of the beam in x-direction. The depth doses and dose profile were normalized to the maximum doses.

Linear accelerator head geometry

Since the geometrical specifications of the Varian VitalBeam linac at ORCI are not available for the previously mentioned reasons, it was difficult to model its actual geometry. To circumvent this limitation, this study used an *ad hoc* (assumed) geometry for all simulations. Simulating this geometry was based on observations that geometrical specifications of the main components of Varian linacs are similar and that the only difference arises from the Phsp parameters of the initial electron beam.[6] For that reason, the main geometrical specifications and data of the Varian VitalBeam linac system used in this study have been defined based on the description made by Bakkali *et al*. 2019. In this study, simulation was performed using G4 linac-MT platform (Geant4-based code with multithreading support). The simulation process was performed in two steps. First, an initial MC simulation of the head components was performed for generating the Phsp by transporting the photons and charged particles from the target to the Phsp defined at 100 cm. Second, the Phsp was then used as source input to the simulation code for performing dose calculation. A screenshot of the head geometry displayed with the OpenGL visualization tool is shown in Figure 1. From the figure, (1) is the target, (2) primary collimators, (3) flattening filters, (4) ionization chamber, (5) Jaws X, (6) Jaws Y and (7) Phsp.

Physics settings

The parameters used in the head simulation and dose calculation are summarized in Table 1. Among several physics lists, electromagnetic standard option 2 (emstandard_opt2) was chosen as suitable physics for simulation. The choice of emstandard_opt2 was based on possible interactions in our simulations. In addition, emstandard_opt2 has been optimized for modeling the transport of photons and charged

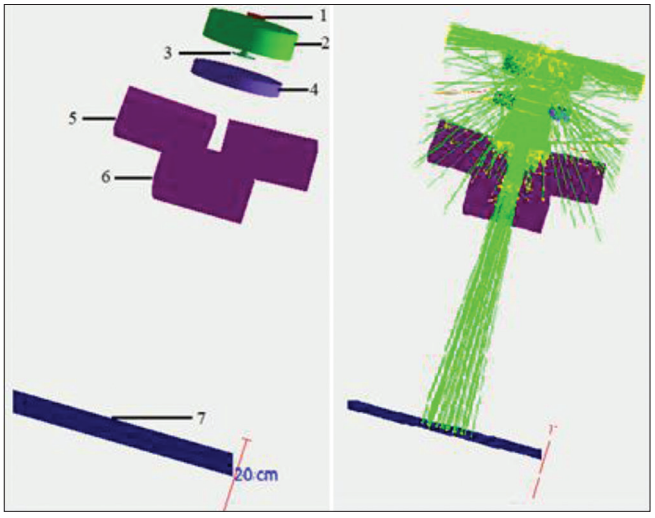

Figure 1: Geometry model of the Varian Linear Accelerator head displayed with OGL visualization tool

Table 1: Parameters used for the linac head simulation and dose calculation

Parameter	Value
Physics List	Emstandard_opt2
Energy cut	60 keV
Production cut	0.1 mm
BREMSPE, split number	80
GAMMATHEC, angle threshold	22

particles for radiation therapy applications.[7] The dose voxel of 0.5 cm × 0.5 cm × 0.5 cm (0.125 cm^3) equals to the active volume of the ionization chamber was used.

To optimize transport parameters, variance reduction techniques (VRTs) were used to reduce dose computation time. Thanks to the G4 linac-MT code that include several VRTs to improve simulation efficiency. In this study, cutoff energy and production cut for secondary particles were used as VRTs. The energy cut for electrons, photons, and gamma were set at 60 keV. When the energy of a particle was below this threshold, the particle was automatically terminated. Only particles with energies above the energy threshold were generated and tracked. The production threshold for electrons, photons, and positron was set at 0.1 mm. The particle recycling was set to 10 so that each particle of the first Phsp is recycled 10 times to increase statistical uncertainty. The other important VRTs used were: Angle cut, which was fixed at 22° so that photons leaving at greater angle than the threshold were killed. The bremsstrahlung splitting was also enabled with NSplit equals to 80.[7]

Dose computations

Before calculating dose distributions in a particular geometry, initial parameters of electron beam must be accurately identified. These parameters are the electron energy and full width at half maximum (FWHM) of both energy and spatial distribution. To identify the appropriate initial parameters of electron beam, three dependable simulations of different configurations of mean energy, sigma and its FWHM were performed.

In the first case, several simulations were performed to adjust the electron beam energy. The first configuration corresponds to initial electron energy of 5.6 MeV, sigma of 0.5 MeV and a 2-D Gaussian distribution in the plane XY with FWHM of 1.18 mm (standard deviation = 0.5 mm). We varied the initial electron energy from 5.6 to 6.4 MeV, with a step size of 0.1 MeV, while sigma of the energy distribution and FWHM of the spatial distribution are fixed at 0.5 MeV and 1.18 mm, respectively. The depth doses were calculated for each energy and compared to the measured doses. The simulations were performed for 10 cm × 10 cm field size in a homogeneous water phantom of 40 cm × 40 cm × 40 cm defined at 100 cm SSD. The phantom geometry was divided into 80 × 80 × 80 voxels in the x, y, and z, respectively. While the x-axis and y-axis were in the cross-plane and in-plane directions, respectively, the z-axis was in the beam direction. The simulations were performed in a personal computer (HP pavilion with a processor Intel® Core™ i7-8550U central processing unit @1.80GHz × 8 and 12 GB RAM installed with Ubuntu 20.04 long-term support operating system). The initial number of simulated electrons was 10^6 and the number of simulated histories was 2.4×10^6. The appropriate initial electron energy was selected by comparing to the best match between the measured and calculated depth doses.

In the second case, after identifying the appropriate initial electron energy that best fit the measured depth doses, the parameters related to the Gaussian energy distribution was optimized by varying the energy sigma considering nine sigma values ranging from 0.1 MeV to 0.9 MeV with a step size of 0.1 MeV. As in the first simulation, the selection of appropriate energy sigma was based on comparison to the best match between measured and calculated depth doses.

In the third set of simulations, the optimal configuration found in the previous cases was taken for fine-tuning the value of the FWHM spatial parameter. The percentage depth dose (PDDs) were calculated for different focal spot sizes. This was performed by varying the focal spot size by considering the following seven sizes: 0.24, 0.47, 0.59, 1.18, 1.41, 1.65, and 1.88 mm, in which the energy and the Gaussian energy distribution of the electron beam are fixed at 6.2 MeV and 0.8 MeV, respectively. The calculations were performed for field size of 10 cm × 10 cm and SSD of 100 cm. As in the first and second simulation, the selection of appropriate of focal spot size was based on comparison between the measured and simulated depth doses. After the appropriate initial electron beam parameters have been determined in the basic simulations, the initial number of simulated electrons was increased from 10^6 to 10^7 initial electrons to increase the statistical uncertainty for the subsequent simulations. Using 10^7 initial electrons, simulations were performed for 10 cm × 10 cm and 20 cm × 20 cm field sizes to validate the beam modeling and comparison of dose distributions.

Data analysis

Since the objective of radiation dosimetry of achieving an accuracy of better than ± 5% for the delivery of dose is desired, gamma index was chosen as the best method. This is because it allows comparison in both dose difference (DD) and distance-to-agreement (DTA) criteria. The gamma index, $\gamma(\vec{r}_m)$ for the gamma function, $\Gamma(r_m, r_s)$ for a measured point at position r_m and simulated point at r_s is calculated using equation (1).[8,9]

$$\gamma(\vec{r}_m) = \min\{\Gamma(\vec{r}_m, \vec{r}_s)\} \forall \{\vec{r}_s\} \quad (1)$$

$$\text{where: } \Gamma(\vec{r}_m, \vec{r}_s) = \sqrt{\frac{r^2(\vec{r}_m, \vec{r}_s)}{\Delta d_M^2} + \frac{\delta^2(\vec{r}_m, \vec{r}_s)}{\Delta D_M^2}} \quad (2)$$

and $r(\vec{r}_m, \vec{r}_s)$ is the distance between $\vec{r}_m$, the position of the measured points, and $\vec{r}_s$, the position of the simulated points, Δd_M is the DTA acceptance criterion and ΔD_M is the dose difference acceptance criterion. $\delta(\vec{r}_m, \vec{r}_s)$ is the difference

between the dose values of the simulated and measured points. The gamma index for each calculated dose point was evaluated for its γ-value to determine if both DD and DTA pass the criterion. If $\gamma(\vec{r}_m) < 1$, the calculated dose point pass the gamma index criteria and if $\gamma(\vec{r}_m) > 1$ fail to pass the gamma index criteria. In this study, the standard acceptance criterion of 2%, 2 mm ($\gamma_{2,2}$) recommended by the American Association of Physicists in Medicine (AAPM) was used.[9] The proper configuration of initial energy, sigma and FWHM that produced the best gamma index result was considered the best configuration for determining the dose distributions.

The index of beam quality described in terms of the tissue phantom ratio, $TPR_{20,10}$ was used to validate the beam modeling as recommended by the TRS-398 dosimetry protocols. The $TPR_{20,10}$ values were obtained from the measured $PDD_{20\,cm}$ and $PDD_{10\,cm}$ data using an approximation equation 3.[10]

$$TPR_{20,10} = 1.2661 \times PDD_{20/10} - 0.0595 \quad (3)$$

Results

Initial electron energy

The measured and calculated depth doses for nine initial electron energies ranging from 5.6 to 06.4 MeV at constant sigma of 0.5 MeV for 10 × 10 cm^2 field size are compared in Figure 2. As expected, the depth of the maximum dose is 1.5 cm for 6 MV photon beam. A closer look in Figure 2 identified that, the initial energy of 6.2 MeV was in good agreement with the measured depth doses. The appropriate energy of 6.2 MeV was justified by the results of the gamma index analysis for each initial electron energy in Table 2. From the table, with exception of the first two dose points at the build-up region, the gamma index-passing rate showed 96% of the calculated depth dose points agree well with measured doses within the acceptance criterion of $\gamma_{2,2}$. Regardless of the discrepancy in the penumbra, it can thus asserted that the initial electron energy of 6.2 MeV is an appropriate for 6 MV photon beam.

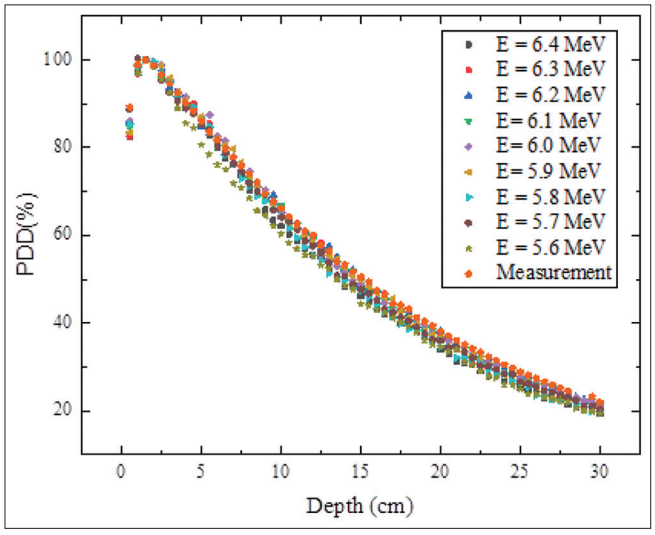

Figure 2: Comparison of percentage depth dose curves calculated for various mean energy of 6 MV photon beam for 10 × 10 cm^2 field size

Energy sigma

The PDD calculated for various energy sigma values and measurements were compared. The results of gamma index tests with acceptance criterion of $\gamma_{2,2}$ for each sigma value are presented in Table 3. The gamma index results showed the best agreement between the measured and simulated PDDs is obtained for 0.8 MeV sigma value. At this sigma value, 98% of the calculated dose points passed the standard acceptance criteria of $\gamma_{2,2}$ (with exception of the first two dose points at the build-up region). From these results, it can be concluded that the initial electron energy and sigma of 6.2 MeV and 0.8 MeV, respectively, best reproduce the depth doses of a 6 MV photon beam produced by the Varian VitalBeam medical system.

Focal spot size

The measured and the calculated PDD obtained for different focal spot size values, in which the energy and sigma are 6.2 MeV and 0.8 MeV, respectively, were compared. The result of gamma index tests for different sigma values are presented in Table 4. From the table, the focal spot size of

Table 2: Gamma index test results for each energy of the electron beam

Gaussian energy parameters		Spatial parameters FWHM (mm)	GI <1%	GI <0.5%
E (MeV)	σ (MeV)			
5.6	0.5	1.18	13.0	8.0
5.7	0.5	1.18	51.0	12.0
5.8	0.5	1.18	29.0	18.0
5.9	0.5	1.18	82.0	41.0
6.0	0.5	1.18	84.0	36.0
6.1	0.5	1.18	92.0	52.0
6.2	0.5	1.18	96.0	66.0
6.3	0.5	1.18	82.0	57.0
6.4	0.5	1.18	24.0	17.0

FWHM: Full width at half maximum, GI: Gamma index

Table 3: Gamma index test results from different sigma value of the electron beam

Gaussian energy parameters		GI <1 (%)	GI <0.5 (%)
E (MeV)	σ (MeV)		
6.2	0.1	6.0	5.0
6.2	0.2	44.0	12.0
6.2	0.3	96.0	73.0
6.2	0.4	60.0	20.0
6.2	0.5	96.0	67.0
6.2	0.6	19.0	3.0
6.2	0.7	94.0	80.0
6.2	0.8	98.0	86.0
6.2	0.9	94.0	63.0

GI: Gamma index

0.5 mm produced the highest agreement with measurements. The gamma-passing rate showed 98% of the calculated dose points were within the standard acceptance criterion of $\gamma_{2,2}$. In view of these results, it can thus be plausible to establish that the electron beam configuration with a mean energy of 6.2 MeV, sigma of 0.8 MeV and FWHM of 1.18 mm, is the appropriate configuration for 6 MV photon beam produced by Varian VitalBeam medical accelerator.

Comparison of depth doses

After the determination of the initial electron beam parameters, the validation of the PDD results of the simulation was performed. Figure 3 shows the comparison of the measured and simulation results of PDD curves for 10 cm × 10 cm and 20 cm × 20 cm field sizes. It is observed that, with exception of the first depth dose point in the build-up region, there was near-perfect agreement between simulated and measured PDDs at two different field sizes. This agreement at two different field sizes suggests the simulated photon beam can cover other field sizes. Implicit in the agreement between simulated and calculated is that, simulation of 6 MV photon beam from Varian VitalBeam linac was successfully achieved. Nevertheless, the dose point at the build-up region showed relatively poor agreement with the measured doses. At this region, dose differences of 4.8% and 4% were observed for field sizes of 10 cm × 10 cm and 20 cm × 20 cm, respectively. However, of particular interest was that 98% of dose points passed the standard acceptance criterion of $\gamma_{2,2}$. With a more stringent criterion of 1%, 1 mm showed that 96% of the calculated dose points matched the measured doses. This observation justifies the success of the simulated 6 MV photon beam. The maximum discrepancy at the build-up is also evident in the gamma index values at each dose point in Figure 4.

Table 4: Gamma index test results for each focal spot size

Focal spot size (sigma X and Y) (mm)	FWHM* (mm)	GI <1 (%)	GI <0.5 (%)
0.1	0.24	92	64
0.2	0.47	94	87
0.3	0.71	39	5
0.4	0.94	96	80
0.5	1.18	98	85
0.6	1.41	98	55
0.7	1.65	92	71
0.8	1.88	72	22

*FWHM=2.355×Sigma. FWHM: Full width at half maximum, GI: Gamma index

The results of gamma index analysis along with the measured and simulated beam quality index values obtained using equation 3 for 10 cm × 10 cm and 20 cm × 20 cm field sizes are summarized in Table 5. It is observed that the calculated $TPR_{20,10}$ for the simulation was found to be 0.668, which is 0.1% higher than the measured $TPR_{20,10}$ which was found to be 0.667. As expected, the $TPR_{20,10}$ increased by about 7% when the field size was increased from 10 cm × 10 cm to 20 cm × 20 cm. It can be concluded that an MC 6 MV photon beam model of the Varian linac was successfully simulated with high accuracy.

Comparison of beam profile

A comparison between the measured and calculated lateral dose profile scored at depth of 10 cm for 10 cm × 10 cm and 20 cm × 20 cm field sizes is presented in Figure 5. Figure 5 shows that the MC-calculated dose points located at the isodose region agreed most closely with the measured doses. Unfortunately, however, this agreement fails at five dose points in the penumbra region with large dose differences of up to 10%. The calculated doses in the penumbra overestimated the measured doses. However, inside the photon field, most of the gamma index values were <1, passing the standard criterion of $\gamma_{2,2}$. In the umbra region, the calculated doses produced reasonably high agreement with the measured doses.

The associated gamma index result at each position from CAX at 10 cm depth for 10 cm × 10 cm and 20 cm × 20 cm field sizes is shown in Figure 6. The results of gamma index tests showed that 66% of the calculated dose points were in

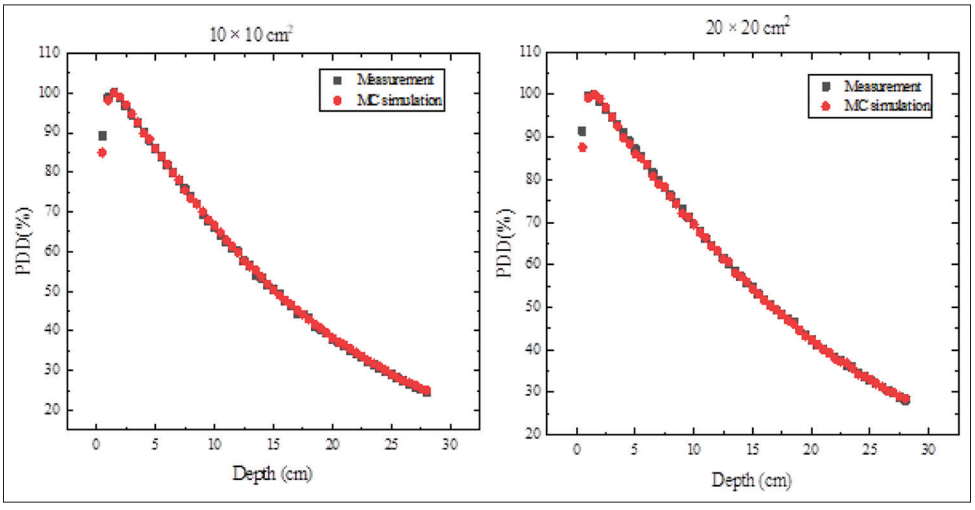

Figure 3: Percentage depth doses of calculated and measured doses for 10 × 10 and 20 × 20 cm² field sizes

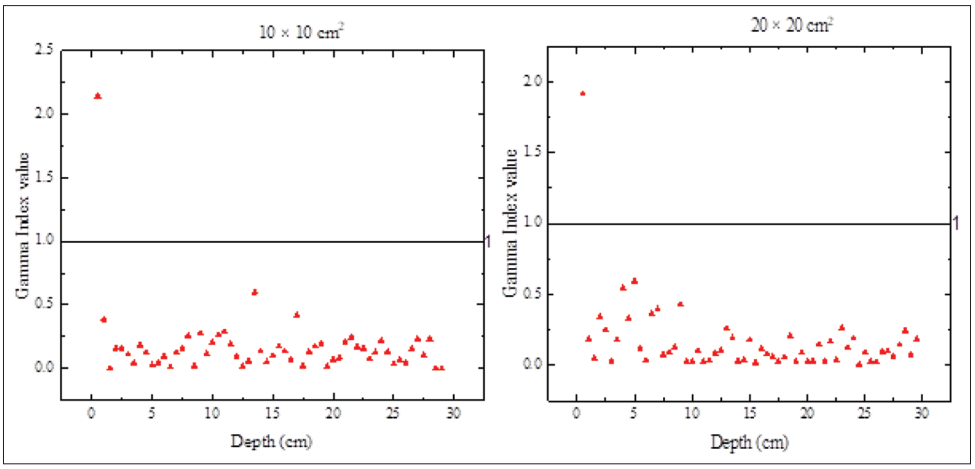

Figure 4: Gamma index value for calculated percentage depth dose for 10 × 10 and 20 × 20 cm² field sizes

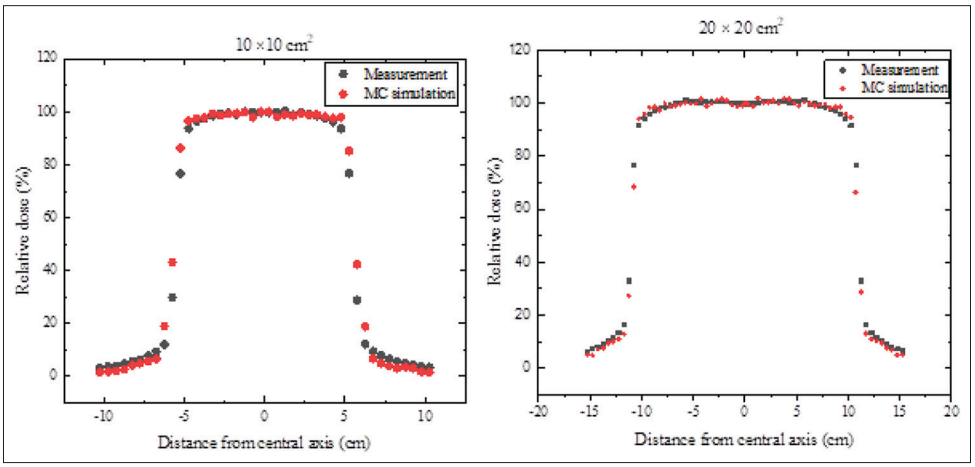

Figure 5: Beam profile of the calculated and measured doses for 10 × 10 and 20 × 20 cm² field sizes

Table 5: Comparison of gamma index test results and $TPR_{20,10}$ for two different field sizes

Field (cm^2)	GI <1 (%)	GI <0.5 (%)	Simulated $TPR_{20,10}$	Measured $TPR_{20,10}$
10×10	98	96	0.668	0.667
20×20	98	95	0.713	0.711

GI: Gamma index, TPR: Tissue phantom ratio

good agreement passing the acceptance criterion of $\gamma_{2,2}$ for 10 cm × 10 cm field size. Luckily, the gamma-passing rate of 78% for 20 cm × 20 cm field size was seemed to be improved. The cause of the discrepancy in gamma-passing rate at two different field sizes could be attributed to the fact that few simulation points were evaluated for smaller field size of 10 cm × 10 cm thus increasing the dose difference compared with the larger field size of 20 cm × 20 cm.

Discussion

In practice, the primary objective of MC simulation in dosimetry is to generate information on distribution and amount of dose, often when measurements are not feasible or difficult to set-up. However, before MC simulation code is used, it must be validated with reference measurements. The objective of this study was to simulate a 6 MV photon beam produced by Varian VitalBeam medical accelerator for further investigations aimed at stimulating clinical applications to improve dose delivery and planning.

The PDDs curves for the calculated and measured dose points in Figure 3 showed excellent agreement with the measurements. The value of $\gamma_{2,2}$ obtained showed 98% of the dose points passing the gamma index test for both field sizes. This accuracy is better than the one reported by previous studies which validated the Geant 4 code for 6 MV photon beam of Varian linacs within 3%, 3 mm accuracy.[11] With the more stringent gamma index criterion of 1%, 1 mm showed 96% of the points passing the gamma index test. This is better than the standard criterion of $\gamma_{2,2}$ proposed by AAPM. This model validation results suggest that the simulation model is accurate and it can be used for the simulation of clinical applications in radiation therapy. Despite this favorable results, the dose point (s) at the build-up region produced large dose difference. This discrepancy in the build-up have also been reported in

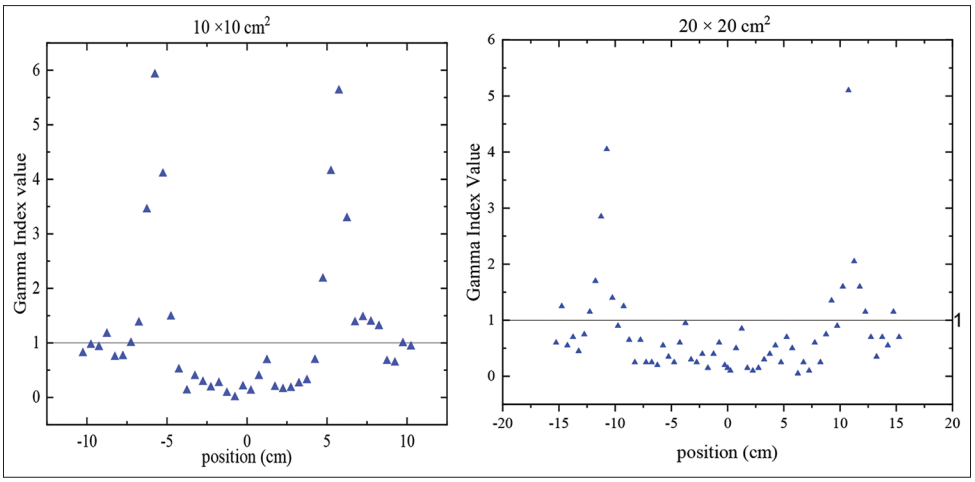

Figure 6: Gamma index value at each position from CAX for calculated dose profile for field sizes of 10 × 10 cm² and 20 × 20 cm²

previous studies.[2,7] The possible reason for the observed discrepancy might be explained by the fact that electronic equilibrium or charged particle equilibrium does not exist at the build-up region.[12] In addition, previous studies outlines that for a good dose measurement in the build-up region, a fine size ionization chamber (parallel plate ionization chambers) is required.[13] Thus, this could be the reason for the large dose differences in the build-up. However, in view of these results and similar studies seem to suggest that accurate estimation/or dosimetry of the build-up dose is important and must be taken into account from the radiation therapy point of view. This is important because, acute skin reaction or delayed effects may occur after overdose or underdose particularly when the skin is part of the tumor volume or the skin is dose-limiting tissue. The authors therefore suggest improvement in the emstandard_opt2 model for accurate build-up dosimetry by G4 linac-MT platform.

The $TPR_{20,10}$ results obtained with simulation confirms that the simulated 6 MV photon beam has similar quality in terms of particle fluence as the actual beam produced by the Varian VitalBeam linac. The discrepancy between measurement and simulation for $TPR_{20,10}$ value was 0.1%. This justifies that the primary source energy was correctly tuned. The observed $TPR_{20,10}$ values are comparable with the reported value in the literature.[14,15]

For dose profile, the dose points in the penumbra produced the highest dose error between the measured and calculated doses. The higher dose error of up to 10% in the penumbra region have also been reported in previous studies.[16,17] The higher level of dose error in the penumbra region may be explained by the presence of high-dose gradient in the penumbra region. The dose errors in the high-dose gradient region increase significantly the overall relative errors between simulation and measurements. Another reason could be inaccuracies in collimation, possible inaccuracies in geometrical specifications and insufficient number of initial particles. As stressed previously, this study used an assumed geometry based on the fact that geometrical specifications of the main components of Varian linacs are similar and that they only differs in the parameters of the initial electron beam. Thus it was difficult to simulate the exact geometry. The authors consider that lack of specifications and hence possible inaccuracies of collimation properties of the jaws components is the likely reason behind the mismatch on the penumbra region. However, despite the discrepancy in the penumbra region, the results for dose profile seem to suggest that there were no significant errors in the simulated photon beam model especially because inside the photon field, most of the gamma index values were <1, passing the standard criterion of $\gamma_{2,2}$. While it was expected that the gamma-passing rate would be more or less the same for different field sizes, the causes of the observed discrepancy in field sizes are not easy to explain.

As reported in the previous studies, precise dose calculation by MC simulation requires sufficiently large number of initial particles in the order of 10^9.[18] Unfortunately, however, following the computational limitation of the resource available, simulation in this study used only 10^7 initial electrons. The estimated simulation time for 10^9 initial electrons is 25 days. Thus, the main shortcomings of the MC technique are the achievable simulation times.

Conclusion

In this study, an MC 6 MV photon beam from VitalBeam Varian linac has been simulated and the dose distributions were benchmarked with measurements for 10 cm × 10 cm and 20 cm × 20 cm field sizes. It was found that best modeling for the photon beam of nominal energy 6 MV from Varian linac corresponds to energy, sigma and FWHM of 6.2 MeV, 0.8 MeV and 1.18 mm, respectively. These initial simulation parameters can reproduce the measured dose distributions, and are expected to reproduce dose distributions for smaller field sizes. For PDD curves, an excellent agreement of 98% between simulation and measured dose distributions was observed, passing the $\gamma_{2,2}$ criteria of the gamma test. However, large dose differences of up to 5% dose at the build-up region

were expected. The index of beam quality, $TPR_{20,10}$ value justified the simulated $TPR_{20,10}$ was in good agreement with the measurements with a discrepancy of 0.1%. This fully justifies the appropriate tuning of the primary source energy. For dose profile, the gamma index analysis showed an acceptable agreement of up to 78% between simulated and measured doses. The low gamma-passing rate for beam profile could be attributed to the mismatch observed in the penumbra region. Despite this discrepancy, most of the calculated dose points inside the photon field agreed closely with the measured doses. In view of these results, the simulation of photon beam for 6 MV Varian VitalBeam linac was successfully with reasonably good accuracy. It can be concluded that the simulated photon beam can be used as radiation source for calculating dose distributions. These results serve as a basis for further investigations aimed at improving dose delivery and planning in cancer patients. However, the authors recommend further validation of dose predictions accuracy with different smaller field sizes and heterogeneous phantom.

Acknowledgment

The authors would like to thank the Ocean Road Cancer Institute for permission to use Varian medical accelerator facility.

Financial support and sponsorship

Nil.

Conflicts of interest

There are no conflicts of interest.

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