Developing a New Monte Carlo Algorithm as an Alternative Tool to Simulate Virtual Source Model on an Elekta Versa Hd Linac

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DEVELOPING A NEW MONTE CARLO ALGORITHM AS AN ALTERNATIVE TOOL TO SIMULATE VIRTUAL SOURCE MODEL ON AN ELEKTA VERSA HD LINAC

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Abstract

The aim of this study is to develop in-house Monte Carlo (MC) algorithm as an alternative tool to simulate all mechanisms of interactions for dose calculation in radiation therapy. Additionally, evaluating the MC-SCAN algorithm with beam modeling of Elekta VersaHD Linac was aimed. Mohan 6 MV photon spectrum was used and the source was modeled as a point source. The absorbed dose is calculated via modeling all interaction mechanisms to gain energy deposition. TPR$_{20/10}$, PDD, photon profile of different field sizes as well as point dose was calculated after that, a comparison of all the achieved results with experimental measurements went through. The difference between calculated mean energy and theoretical one was obtained $<1\%$. The calculated TPR$_{20/10}$ value was $(0.685 \pm 0.03)$. The difference between experimental PDD and photon profile, and the calculated ones by new model was $<3\%$ for all field sizes. A new MC algorithm can be used as an alternative tool to simulate virtual source model on an Elekta VersaHD Linac. All calculated values turned out to present the difference between experimental measurements to be $<3\%$ for homogeneous mediums. Nonetheless, further research was needed to be done for the better improvement.

Key words: Beam model, Monte Carlo simulation, Radiation dose, Radiation Therapy
1. Introduction

Calculating the required radiation dose is significantly highlighted in radiation therapy. For the same reason, in ICRU Report-24 it was suggested that, the difference between the absorbed dose and the planned prescribed dose should be within \( \pm 5\% \) for the suggested degree of accuracy for the delivery of treatment in radiation therapy (1, 2). Every step of dose delivery - i.e. acquisition of patient-specific tumor information, dose calculation, machine calibration and patient positioning - is required to be performed with a level of accuracy higher than 5\% for which to be achieved both detailed and accurate anatomical information as well as rather rigorous and accurate dose calculation algorithm are required (3, 4). There are two classifications namely measurement-based and model-based approaches for dose calculation algorithm in TPS. For the most of the model-based algorithm, a kernel, which contains pre-calculated data via Monte Carlo, describes the energy spread by secondary particles (5-7).

Theoretically, Monte Carlo (MC) method offers the most complete and exact system of dose calculation benefitting from true 3D geometry as simulation of tracks of individual particles (8, 9). In the sampling process; appropriate quantities of computer-generated pseudo-random numbers are used to simulate ionizing radiation transport. Additionally, tracks of individual particles are simulated by the method from the probability distributions governing the individual physical process. On the other side, a large batch of particles are tracked to calculate average values of macroscopic quantities, such as particle fluence, energy spectrum and absorbed dose distribution (10).

Monte Carlo based treatment planning systems (TPS) need information regarding the radiation beam incident on the patient in detail for accurate dose calculation (11). These information is called phase space (PS) data which includes minimum the energy, position, direction and type (electron, photon, and positron) of radiation particles directed toward the patient. The PS is needed to initiate transport of particles at the patient surface. The transport process of such particles are simulated to determine the absorbed dose in the medium (12, 13). However, such information is too difficult to be achieved through direct measurements for a clinical beam. The intensities of high radiation at the LINAC exit window as well as the availability of suitable detectors are the limiting factors to perform direct measurements of PS data. Additionally, the storage requirement should be up to tens of gigabytes for the phase space files containing millions of photons. On the other
hand, beam modeling is an alternative for the direct input to obtain the required PS information and to reduce considerably on both computing time and storage (14). For a full MC simulation of LINAC head information that includes a variety of factors, i.e. geometry of the treatment head, source position, energy and angular distribution, dosimetric measurements consist of square/rectangular open field data, percent depth dose, profiles and output factors are needed to be performed (15, 16). For the MC dose calculation, the beams reconstructed from these models are used as the radiation source which is produced by linear accelerators. Comparing the calculated results with the dosimetric measurements should be performed for the model evaluation (17-19).

In the scientific societies, it is accepted that MC simulation can accurately model the physical process which occur during the particle transport in media, it’s popularity in the field of radiation therapy gained momentum since the 1970s. A number of MC-based simulation codes, e.g., MCNP, Geant4, PENELLOPE, EGSnrc, are currently in use for radiation transport simulation. These MC simulation codes plays a crucial role for accurate patient-specific dose calculation in radiation therapy (20, 21).

Most of the commercial code like BEAM/EGSnrc allows users to easily congregate a LINAC models however, the target modeling is the most important part in this model. Beam characteristics such as energy, angular and spatial distribution of the particles in the beam are mandatory requirement for MC method (22). On the same ground, efforts to improve these models are being made. On the other hand, the long computation time makes MC methods often impractical for routine clinic applications (23). To solve this problem, many research focused on to reduce computation time by using different variance reduction technique (VRT). Garcia-Pareja et al. (24) compared multiple VRTs and evaluate their performance. In other paper by Samir Didi et al. (25) simulated Elekta Synergy LINAC using Geant4 Monte Carlo code under grid computing environment to reproduce therapeutic dose distributions in a water phantom with an accuracy less than 2%.

This presented study aimed at developing an in-house general-purpose Monte Carlo dose calculation algorithm namely, MC-SCAN, as an alternative tool to simulate particle transport. Additionally, it was aimed to simulate all interaction mechanisms of dose calculation in radiation therapy and to evaluate MC-SCAN dose engine through beam modeling of Elekta VersaHD linear accelerator. To do this simulation binning approach and was used and target was modeled as a
point source. To reduce the statistical uncertainty and increase the accuracy, we tracked $10^{10}$ photon for dose calculation. Moreover, Mersenne Twister algorithm was implemented to generate random number. Additionally, the MC-SCAN Monte Carlo code was tested for many parameters including tissue phantom ratio, percent depth dose, beam profile and point dose in uniform water phantom and RW3 solid phantom.
2. Materials and Methods

2.1. Experimental measurements

Instructions provided by the manufacturer as well as AAPM TG-106 are the sources of beam data collection in Monaco TPS commissioning. All point dose, percent depth dose (PDD) and beam profile measurements were performed with a PTW BEAMSCAN Water Phantom (PTW, Freiburg, Germany), equipped with a PMMA water tank. All PDDs and as well as profile measurements were obtained at 90 cm source to surface distance (SSD) for field size from 5x5 cm² to 40x40 cm². beam profile measurements contained in-plane, cross-plane, and diagonal-plane was obtained at 10 cm depth. PTW’s MEPHYSTO Navigator software was used for the purpose of processing scanned data. Additionally, all point dose measurements in water were performed at 90 cm SSD and at the depth of 10 cm. The PTW Semiflex 3D chamber (active volume = 0.07 cm³) was used for the all field sizes. Additionally, absolute dose calibration, TPR20/10 and point dose measurements happened through PTW RW3 phantom and PTW Farmer ion chamber (active volume = 0.6 cm³).

2.2. MC-SCAN Modeling

Mersenne Twister (MT) algorithm was used to generate random number in MC-SCAN modeling which presents us with a period \(2^{19937} - 1\) along with 623 dimensional equidistributional up to 32-bit accuracy, while using a working area of only 624 words.

Beam modeling for photon beams was illustrated in Figure-1. As shown, one phase space plane, which was for verification of the source model when dose distributions in medium were calculated, was designed. The origin of the coordinate system \((x,y,z) = (0, 0, 0)\) is at the front surface of the target where the photons are incident. \((x,y,z) = (0, 0, 100)\) cm was the isocenter of the machine. Phase space plane was at 100 cm at the target. In MC-SCAN modelling the binning approach was used for beam characteristics in order to find a function to fit the beam throughout phase space. The binning approaches needed limited knowledge about the beam characteristics that is easily acquired, such as the field size and maximum energy. Mohan 6 MV photon spectrum was used in MC-SCAN modeling to simulate photon fluence and angular distribution. The boundary conditions were taken as \(a = 0, b = E_{\text{max}} = 6\text{MeV}\). The \(f(E)\) function developed by Ozmutlu (26, 27) was used for the fit function of the Mohan Spectrum.

\[
f(E) = P_1 \times e^{-(E+P_2)^2/(2P_3^2)} + P_4 \times E \times e^{-E^2/(2P_5^2)}
\]  

(1)
In the function Ps are as follows: \( P_1 = 1.488011, P_2 = 3.621140, P_3 = 3.086142, P_4 = -3.804164, P_5 = 0.09053989 \).

PTW BeamScan water phantom (dimension = 50x50x50 cm\(^3\)) and PTW RW3 phantom (dimension = 30x30x30 cm\(^3\)) geometry were taken advantage of in MC-SCAN modeling. In order to calculate the absorbed dose within the respective phantom geometries, the phantoms were divided into 5 mm voxels. The geometry of the phantom simulated via MC-SCAN was presented in Figure-2. Since the source dimensions are very small compared to the source to surface distance (SSD), the source is modeled as a point source. It was assumed that, the probability of a \( \gamma \)-ray from an isotropic point source is the same for all directions. The direction of a \( \gamma \)-ray emitted from the source was determined in spherical coordinates by the polar angle (\( \theta \)) and the azimuth angle (\( \phi \)). The polar angle of the direction of the \( \gamma \)-ray emitted from the point source varies in the range of \( 0 \leq \theta \leq \pi \), and the azimuth angle varies in the range of \( 0 \leq \phi \leq 2\pi \). In MC-SCAN dose engine, the \( \gamma \)-ray directions were determined randomly. In the modeling, the A-P direction was accepted as the x-axis, the gun-target direction as the y-axis, and the target-phantom direction, that is, the SSD direction, as the z-axis. The point source – phantom geometry and representation of photons in MC-SCAN modeling are shown in Figure-3. Rotation matrix \( R(\theta, \phi) \) was determined in terms of pole angle and azimuth angle with following equation (28, 29).

\[
R_z(\phi)R_y(\theta)\hat{z} = \begin{pmatrix}
sin\theta \cos\phi \\
sin\theta \sin\phi \\
cos\theta 
\end{pmatrix}
\]

(2)

Passing within the phantom in a certain distance with no interaction, the photon loses energy and intensity (I) after a certain distance; this decrease was presented by the following equation (30).

\[
I = I_0 e^{-\mu x}
\]

(3)

After entering the medium, the first interaction of the photon was determined by the shortest of the mean free paths. Thus, both the mean free path of each photon and the type of interactions were determined. Provided the interaction type is Rayleigh scattering, neither atomic excitation nor energy transfer to the environment happened, however, all interaction types were included in the MC-SCAN modeling. The cross section of the interaction in Rayleigh scattering was modeled by the Thompson Scattering equation as follows:
\[
\frac{d\sigma_{\text{Koh}}}{d\theta} = \frac{r_e^2}{2} \left( 1 + \cos^2 \theta \right) F^2(x, Z)
\]  

(4)

\(F(x, Z)\) is the Thompson scattering function of a single molecule of the medium and is expressed by the following equation. For 6 MV photon energy, all \(F(x, Z)\) values between 1 keV - 6 MeV were calculated by the linear interpolation method (31).

\[
F^2(x, Z) = \sum_{i=1}^{N} n_i F_i^2
\]

(5)

Provided the interaction type had been photoelectric effect, it had been assumed that the range of electrons formed in the photoelectric effect would have been very small compared to the phantom size. Additionally, the incident photon transferred its energy to the medium and the coordinates of the absorbed energy did not change. For the same reason, once the interaction type was photoelectric effect, the energy of the incident photon was assumed to be absorbed at the interaction point in MC-SCAN modeling and a new photon tracking was started.

In the modeling of Compton scattering, the total cross section for the interaction with atomic electrons of the incident photon, owning \(E\) energy, was modeled with the Klein-Nishina equation.

\[
\frac{d\sigma_{\text{KN}}}{d\theta} = \frac{r_e^2}{2} \left( \frac{E'}{E} \right)^2 \left( \frac{E'}{E} + \frac{E}{E'} - \sin^2 \theta \right) S(x, Z)
\]

(6)

\(S(x, Z)\) is the Compton scattering function for a molecule of the scattering medium and is expressed as follows. For 6 MV photon energy, all \(S(x, Z)\) values between 1 keV – 6 MeV were calculated by linear interpolation (31, 32).

\[
S(x, Z) = \sum_{i=1}^{N} n_i S_i (v, Z_i)
\]

(7)

In the equation, \(r_e\) is \(e^2/mc^2\) and it is the classical electron radius. In Compton scattering, the electron energy escapes from the voxels was neglected and it was assumed that the energy of the electrons was absorbed at the point where Compton scattering occurred. From this moment, photon tracking continued with the scattered photon. The semi-empirical model was used to model pair production for an atom with \(Z\) atomic number. In MC-SCAN modeling, the total cross section of pair production was modeled based on Bethe-Heitler formula (28, 33).

\[
\frac{d\sigma_{\text{pp}}}{d\epsilon} = r_e^2 \alpha Z [Z + \eta] C_r \frac{2}{3} \left[ 2 \left( \frac{1}{2} - \epsilon \right)^2 \phi_1(\epsilon) + \phi_2(\epsilon) \right]
\]

(8)
In the formula, \( C_r \) is the radiative correction coefficient and its value is 1.0093. for 6 MV photon energy, triplet formation is neglected and \( \epsilon \) is the reduced energy of the electron calculated with the below-mentioned equation. The reduced energy (\( \epsilon \)) value of the electron was accepted to vary between \( k-1, 1-(k-1) \) where \( k = \frac{E}{m_e c^2} \).

\[
\epsilon = \left( \frac{E + m_e c^2}{E} \right)
\]

(9)

The angular distribution of the produced particles was not derived solely from the kinetic energy of the particle. The polar angles \( (\theta_-, \theta_+) \) of the motion directions of the electron and positron relative to the direction of the incident photon were simulated from the expression obtained in high energy theory.

\[
p(\cos \theta_\pm) = \alpha (1 - \beta_\pm \cos \theta_\pm)^{-2}
\]

(10)

\[
\beta_\pm = \sqrt{\frac{E_\pm (E_\pm + 2m_e c^2)}{E_\pm + m_e c^2}}
\]

(11)

In the equation, \( \alpha \) is the normalization constant and its value is 1/137.035. The particle angle \( \cos \theta_\pm \) was randomly sampled according to the following equation.

\[
\cos \theta_\pm = \frac{2\xi-1+\beta_\pm}{(2\xi-1)\beta_\pm+1}
\]

(12)

The directions of the produced particles and the incident photon do not have to be co-planar. The azimuth angles of electron and positron \( (\theta_-, \theta_+) \) were modeled independently and assumed to vary between \( 0-2\pi \) (28). This point was determined as the point where the electron and positron pair annihilated and annihilation photon was emitted. From this moment on, the annihilation photon was tracked. The photon tracking logic of MC-SCAN modeling is shown in Figure-4. In MC-SCAN modeling approximately \( 10^{10} \) photon were tracked to reduce the statistical uncertainty by increasing the history of particles. Of all the interaction mechanisms, the photon energy was saved. Then, the absorbed dose was calculated based on the following equation.

\[
Dose = \frac{\text{Energy}}{\text{Mass}}
\]

(13)
2.3. Beam modeling with MC-SCAN dose engine

Beam model simulation of Elekta VersaHD LINAC was performed through MC-SCAN dose engine for accurate dose calculation. For phantom scatter factors ($S_c$), water phantom and RW3 phantom were modeled. $S_c$ values were calculated for all field sizes from $2 \times 2 \text{ cm}^2$ to $40 \times 40 \text{ cm}^2$. Then, phantom scattering factors were calculated by normalizing the dose values obtained in other fields to the reference field $10 \times 10 \text{ cm}^2$. The calculated factors were compared to the experimental ones. PTW RW3 phantom geometry was modeled to calculated tissue phantom ratio (TPR$_{20/10}$) for $10 \times 10 \text{ cm}^2$ field size. The absorbed dose was calculated for 10 cm depth at SSD 90 cm and 20 cm depth at SSD 80 cm. And then calculated TPR$_{20/10}$ was compared with measured one. Water phantom geometry was modeled to simulate percent depth dose (PDD) and photon profile for fields from $5 \times 5 \text{ cm}^2$ to $40 \times 40 \text{ cm}^2$. For PDD simulations, the absorbed dose was calculated for every 5 mm, starting at a depth of 35 cm on the z-axis. In all profile simulations, the absorbed dose values were calculated at 5 mm intervals in the x-direction (Crossplane), starting from 3 cm outside the starting point of the aforementioned fields. The absorbed dose values obtained at each distance were normalized to the dose obtained at the baseline on the x-axis. The flatness and symmetry was calculated with following equations.

$$Flatness = \frac{D_{\text{max}}}{D_{\text{min}}} \times 100$$

(14)

$$Symmetry = \frac{(D_{\text{max}} - D_{\text{min}})}{(D_{\text{max}} + D_{\text{min}})} \times 100$$

(15)

For photon profile, SSD was 90 cm and depth was 10 cm to eliminate the effect of electron contamination. The calculated PDD and photon profile were compared experimental measurements. Additionally, paired sampled t-test was performed to determine the difference between those values statistically significant. To perform point dose measurement, a treatment plan was created for $10 \times 10 \text{ cm}^2$ open field. Final dose calculation was done with Monte Carlo algorithm using Monaco Treatment Planning System v.5.51. The planned dose was compared with absorbed dose and calculated dose via MC-SCAN dose engine.
3. Results

3.1. Energy Spectrum

Electrons accelerated in clinical linear accelerators produce high-energy beams after hitting the tungsten target. The maximum energy of photons formed by electrons accelerated with 6MeV energy is 6 MV. The bremsstrahlung spectrum of photons ranges from 0 keV - 6MeV. The 6 MeV photon spectrum was simulated through MC-SCAN dose calculation algorithm. The simulated spectrum was shown in Figure-5. The mean photon energy was approximately 1.983 MV. The difference between calculated mean energy and theoretical one (2 MV) is <1%. The values obtained by MC-SCAN modeling provided the manufacturer’s criterion of <3% for 6 MV photon energy spectrum.

3.2. Phantom Scatter Factors

Output factors expressing the phantom scattering factor for Monte Carlo based treatment planning systems were obtained by normalizing the values obtained in different areas to the values obtained in different areas to the values obtained in the reference area. The output factors obtained in water phantom and PTW RW3 phantom were compared with the factors obtained by MC-SCAN modeling. Phantom scatter factors for different field size were listed in Table 1 and Table 2. The difference between experimental measurements and MC-SCAN modeling for output factors obtained with the PTW RW3 phantom was >3% for the 2x2 cm$^2$ field size, while they were <3% for the other field sizes. Accordingly, MC-SCAN modeling provided the manufacturer’s criterion of <3% for the output factors obtained in the PTW RW3 phantom for other field sizes except the 2x2 cm$^2$ area.

3.3. Tissue – Phantom Ratio (TPR20/10)

Tissue – Phantom ratio (TPR20/10) was compared as experimental and theoretical value. According to the manufacturer’s criteria; the TPR20/10 value for 6 MV photon energy was determined as 0.67. The experimental TPR20/10 value was measured as (0.676 ± 0.04) and the difference between the theoretical value and the experimental value was <1%. In the second step, TPR20/10 value was calculated through MC-SCAN dose calculation engine and compared with the experimental measurements and theoretical value. The calculated TPR20/10 value via MC-SCAN modeling as (0.685 ± 0.03). The difference between the calculated and measured value was 1.31%.
moreover, the difference between the value calculated value and theoretical value was 2.53%. Thus, the MC-SCAN modeling provides the manufacturer’s criterion of <3% for the TPR_{20/10} value.

### 3.4. Percent Depth Dose (PDD)

For different field sizes, calculated percent depth dose (PDD) through MC-SCAN modeling was compared with experimental percent depth dose. Paired sampled t-test was performed to determine whether there was a statistical difference between both PDD values. The comparison of MC-SCAN modeling and experimental measurements for PDD and the statistical analysis were shown in Table-3. As a result of the analysis; the difference between experimental PDD and PDD calculated by MC-SCAN modeling was <3% for all field sizes. MC-SCAN modeling provided the manufacturer’s criterion of <3% for PDD value for all fields. According to the statistical analysis results, although there was a significant difference (p = 0.00) between MC-SCAN modeling and experimental measurements in the PDD value obtained for the 10x10 cm² field size, there was no significant difference (p > 0.05) between the two PDD values for the other areas. Calculated and measured PDD’s for different field size were shown in Figure-6.

### 3.5. Photon Profile

Dose profile measurements were obtained for open field sizes at SSD 90 cm. The dose profile was calculated with MC-SCAN dose engine for different field sizes. Paired Sampled t-Test was applied to determine whether there was a statistical difference between experimental measurements and MC-SCAN modeling. The statistical analysis result was shown in Table-4. As a result of the analysis; the difference between the experimental and calculated dose profile was <3% for all field sizes. Thus, the MC-SCAN modeling provided the manufacturer’s criterion of <3% for the photon profile. According to the statistical analysis, the difference between calculated and experimental photon profile was not statistically significant (p > 0.05) for all field sizes. The experimental measurement and MC-SCAN modeling comparison of the dose profile values obtained for different field sizes was shown in Figure-7. At the same time, flatness and symmetry values were calculated from the obtained photon profiles. The difference between experimental measurements and MC-SCAN modeling for flatness and symmetry was shown in Table-5.
3.6. Point Dose

To perform point dose measurement, a treatment plan was created for 10x10 cm$^2$ open field. Final dose calculation was done with Monte Carlo algorithm using Monaco Treatment Planning System v.5.51. The planned dose was $(80.5 \pm 0.05)$ cGy. The measured dose was $(79.9 \pm 0.07)$ cGy. The difference between planned dose and measured one was <1%. The calculated dose through MC-SCAN dose calculation algorithm was $(78.2 \pm 0.05)$ cGy. The difference between calculated dose and planned dose was 2.8%, while the difference between calculated dose and measured dose was 2.15%.
4. Discussion

Considered among the most accurate methods of calculating radiation dose in the radiotherapy, Monte Carlo (MC) simulation is currently a common method for benchmarking the other dose calculation methods in homogeneous and inhomogeneous cases. Some MC codes, including EGSnrc, MCSP, MCNP, Penelope, Geant4 and etc., are widely used in medical physics each coming with a number of pros and cons related to the simulation time as well as output of physical factors obtained at the end of the simulation (18, 34). Every time experimental measurements are not possible, simulation studies are critically taken advantage of so that the physical factors are understood. Hence, the first and the most crucial step in Monte Carlo dose calculations in either phantoms or humans is developing an MC beam model for linear accelerators, however, since the possibility of running into errors is always a possibility, dose calculation error must be calculated in materials and geometry selection when building LINAC (35, 36). Each LINAC modeling using MC code comes with different characteristics resulting in different PDD, dose profile and output factors obtained. The given output at the end of the simulation is affected by a number of simulation parameters, such as voxel sizes, range rejection methods and particles cutoff energy. There are a large number of studies related to modeling LINAC as well as characterizing the LINAC output (16, 37).

In a paper by Samir Didi et al. Elekta Synergy LINAC was simulated via Geant4 Monte Carlo code under a grid computing environment. In the simulation therapeutic dose distribution was reproduced in a water phantom with an accuracy less than 2%. Dose distribution was performed for 6 MV photon energy with different field sizes at a SSD of 100 cm. the difference between the simulation results and the measured data was within the 2%. Gamma index comparisons was over 98% of the points for all simulations (25).

18 MV photon beam was simulated using GATE v8.0 built with GEANT4 v10.3.0 and delivered by Varian Clinac 2100C LINAC. Water phantom was taken advantage of in order to measure PDDs and dose profiles at SSD 100 cm over 4x4 cm² and 10x10 cm² field sizes. Calculating MC was performed assuming the initial conditions so as to reproduce the measurements. Additionally, the dose Actor tool provided by GATE was used to calculate dose distributions. Considered for all fields, M. Fiak et al. argued that more than 96% of the calculated data points passed the Gamma
Index test criterion. Additionally, the difference between the measured and calculated TPR\textsubscript{20/10} was obtained to be <1% (38).

Benefiting from the PRIMO code, Hussin Aamri et al. simulated a 6 MV flattening filter free (FFF) photon beam of a TrueBeam Varian LINAC. PRIMO simulations were used to determine the depth-dose profiles of various jaw open fields as well as cross-beam profiles for various depths inside water phantoms after which were compared with EGSnrc simulation results. The score of 93% match is generally obtained in both codes with a systematic variation in agreement between the two models. However, the closest agreement, within 94%, was obtained for a 6x6 cm\textsuperscript{2} jaw open field at 5.0 cm and 10.0 cm depth (39).

Sitti Yani et al. used the EGSnrc/BEAMnrc-DOSXYZnrc code for the sake of modeling and characterizing the 6 MV and 10 MV photon energy which were generated by Elekta Infinity LINAC. PDD simulation was performed and dose calculation was done for output factors. All results were compared with the experimental measurements; nevertheless, the commissioning process was only conducted for a 10x10 cm2 field size. Both energy levels show the difference between local dose of PDDs to be less than 2%. The output factors measured with different detectors are in good agreement with simulation within less than 2% for field sizes (40).

To calculate initial beam characteristics of a 6 MV FFF photon energy was simulation via The Tool for Particle Simulation (TOPAS) Monte Carlo code. For the simulation Elekta VersaHD LINAC was modeled with the all main component including LINAC head and MLCs. PDDs, beam profiles and TPR\textsubscript{20/10} was calculated in uniform water phantom for the 5x5 cm\textsuperscript{2}, 10x10 cm\textsuperscript{2} and 20x20 cm\textsuperscript{2} field sizes. Based on obtained results, the simulated and measured data were in good agreement, with gamma rate analysis of 3%/3 mm criteria reaching 95% and 99 % for PDD and beam profile respectively (41).

On the other hand, this study offers in-house Monte Carlo dose calculation algorithm, called MC-SCAN, to simulate beam model of Elekta Versa HD LINAC. PTW BeamScan water phantom and PTW RW3 phantom were taken into consideration for beam modeling while voxel size was assumed to be 5 mms in MC-SCAN modeling. Additionally, Mersenne Twister (MT) algorithm was implemented to generate large supplies of random numbers. Deposited energy within the phantom geometry was determined by all interaction mechanisms. Phantom scatter factors for both phantoms, TPR\textsubscript{20/10}, PDD, photon profile for different field sizes (5x5 cm\textsuperscript{2} to 40x40 cm\textsuperscript{2}) and point
dose for 10x10 cm² were simulated via MC-SCAN dose engine and then all results were compared with experimental measurements. In order to reduce the statistical uncertainty in MC-SCAN modeling, approximately $10^{10}$ photon were tracked; hence, the history of particles increased. The results prove that the difference between the calculated values and the measured ones is <3% and MC-SCAN modeling showed good agreement with the experimental measurement for all parameters. MC SCAN being a CPU-based kind of algorithm, a work station with an Intel i9 processor with a processor speed of 3.8 GHz and a 1 TB SSD was used to run the MC-SCAN dose engine. Since all atom-photon interactions were modeled and the number of photons followed was increased, the computation time varied depending on the parameters calculated in MC-SCAN modeling for 10 iterations. Considering a single open space; the simulation was completed between 2 hours and 72 hours for each iteration and each depth. One point needs to be mentioned is that MC-SCAN modeling was tested only on water phantom and RW3 phantom yet complex geometries, such as a human body, was not taken into consideration. Moreover, LINAC head was not simulated in the modeling as the source is considered as a point source since the source dimensions are very small compared to the source to surface distance (SSD). Additionally, comparing MC-SCAN with general purpose Monte Carlo codes such as PENEOPE, Geant4 would be crucial to validate MC-SCAN.
5. Conclusion

The MC-SCAN Monte Carlo engine had the potential to be used as an alternative for simulating beam model on Elekta Versa HD LINAC. The difference between experimental measurements of all calculated values via MC-SCAN was within < 3%. Additionally, MC-SCAN dose engine provided the required degree of accuracy to deliver of the tumor dose which is suggested to be within ±5% by ICRU Report-24. On the other hand, simulation was performed in only homogenous media. Hence, the algorithm needed to have further investigation for better improvement.
6. Limitations

One limitation is that MC-SCAN modeling was tested only on water phantom and RW3 phantom yet complex geometries, such as a human body, was not taken into consideration. Moreover, LINAC head was not simulated in the modeling as the source is considered as a point source since the source dimensions are very small compared to the source to surface distance (SSD). Additionally, comparing MC-SCAN with general purpose Monte Carlo codes such as PENELOPE, Geant4 would be crucial to validate MC-SCAN.
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Author Contributions

Concept-SC, EŞ, DK, NM;

Supervision-DK, NM, EŞ;

Materials and Methods – SC, DK, NM;

Data Collection/Processing – SC;

Analysis and Interpretation – SC, DK, NM;

Writing - SC
Conflict of Interest

No potential conflict of interest was reported by the authors.
Ethical Approval
Not applicable for this study
Informed Consent Form

Not applicable for this study
Funding
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**Data availability statement**

The dataset(s) supporting the conclusions of this article is(are) included within the article (and its additional file(s)).
Consent to Publish

Not applicable for this study
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Figure 1

Beam Modeling for photon beams
Figure 2

Phantom geometry simulated via MC-SCAN modeling
Figure 3

The point source – phantom geometry and representation of photons in MC-SCAN modeling
Figure 4

The photon tracking logic of MC-SCAN modeling
Figure 5

The spectrum of 6 MV Photon energy simulated via MC-SCAN dose calculation algorithm
Figure 6

The comparison between experimental measurements and MC-SCAN calculations for PDD
Figure 7

The comparison between experimental measurements and MC-SCAN calculations for photon profile