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Validation of Varian Clinac iX Model on 6 MV Photon Beam Using Fast Monte Carlo Simulation

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ABSTRACT

Monte Carlo (MC) is widely recognized as the most accurate method for dosimetry analysis in radiotherapy due to its precision. However, successful MC dose calculation hinges upon the validation of the linac model employed in simulations. This study aims to verify the PRIMO model of the Varian Clinac iX and to determine the optimal initial electron energy. The comparison of one-dimensional dose distribution between simulations and measurements serves as the foundation for assessment. The Varian Clinac iX on 6 MV photon beam was meticulously modeled with the initial electron energies spanned from 5.2 to 5.8 MeV in increments of 0.2 MeV. The dose calculation were performed for a field size of 10 cm × 10 cm and a source-to-surface distance (SSD) of 100 cm. The Dose Planning Method (DPM) was adopted as the simulation engine for expedited MC simulation. A number of particle histories–approximately 4.0×10^8–were simulated, resulting in the generation of around 10^9 particles from the linac head. The investigation revealed that an initial electron energy of 5.8 MeV achieves good agreement with measurement by attaining the smallest difference in percentage depth dose (PDD) of about 0.98%. The lateral dose deviation of approximately 4.63% serves to validate the precision of the secondary collimator design. Additionally, a comparative analysis of DPM and PENELOPE for dose calculation was conducted. In contrast to the PENELOPE, the DPM speeds up simulation time by approximately 3.5 times, reduced statistical uncertainties to 0.59% and afford better accuracy in dose calculation. The result underscore the suitability of the PRIMO model for MC simulation for dose calculation, given its robust agreement with the measurements.

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1. INTRODUCTION

The pivotal role of linear accelerators in cancer cell treatment within radiotherapy has been extensively acknowledge [1]-[4]. Accurate estimation of absorbed dose can be achieved through Monte Carlo (MC) calculations, which stand as a preeminent method for radiation transport simulations and precise dose computations in radiotherapy [5]-[18]. The MC method dissects the microscopic interactions within phantoms by analyzing the trajectory of both primary and secondary simulated particles. The use of MC is supported by the fact that, while analytical algorithms employed in treatment planning are suitable for computations in homogeneous phantom, they tend to yield significant inaccuracies when working with heterogeneous material. In contrast, Monte Carlo simulations are unaffected by the medium's composition or density and consistently yield more accurate dose distributions compared to analytical-based calculations [5], [6]. Various MC codes have emerged, with PRIMO standing out as a software tool tailored for MC simulation, using Varian and Elekta clinical linear accelerators as the beam sources [19]-[22].

This versatile software performs dose computation within both phantom and patient computed tomography scans [23], [24] and enabling independent dose calculation to verify the calculations of treatment planning system (TPS). However, to ensure the fidelity of PRIMO's linac model, validation becomes imperative. The validations were done by assessing the dosimetric parameters by comparing measurements at a standard reference field size—often 10 cm × 10 cm—with the corresponding MC [25]-[27]. The essence of linac validation lies in confirming that the beam quality utilized in MC simulation emulates the actual linac's characteristics, given the inherent variations across different linear accelerator configurations. The precision of dose calculation in TPS depends on the accuracy of the linac model and initial beam paramaters, which affects the accomplishment of radiotherapy [8], [27]. In the context of MC simulation, PRIMO extends its versatility by accommodating the PENELOPE and Dose Planning Method (DPM) as the computation engines [24], [28].

PENELOPE effectively models the transport of coupled electron-photon within energy ranges up to 1 GeV, delineating dose distribution in phantoms via the ratio of energy deposited within the region of interest to its mass [29], [30]. DPM, on the other hand, expediently calculates dose for electron and photon transport through optimal utilization of computer CPU resources, a feature that has been proven to substantially reduce computational time in previous studies. DPM employs streamlined cross-sectional models tailored to the energy range utilized in standard radiotherapy and suitable for materials with low atomic numbers, akin to those present within the patient's body [24]. Against this backdrop, The primary objective of this study materializes as the validation of PRIMO's model for the Varian Clinac iX. Additionally, the study aims to quantitatively assess the accuracy of dose calculations through a comparative analysis between DPM and PENELOPE algorithm within a homogeneous phantom scenario.

2. METHODS

2.1. Monte Carlo Simulation

The MC simulations were performed using PRIMO version 0.1.5.1307 on a computer equipped with an Intel i9-11900K processor running at @3.50 GHz, with 16 threads and 8 GB RAM. The beam source for this study was the Varian Clinac 2100 model. The PRIMO workflow is illustrated in Fig. 1.

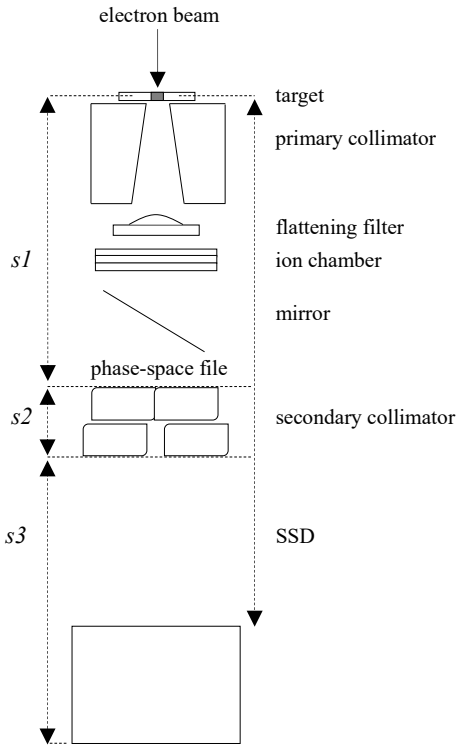


Fig. 1. Simulation Segmentation of Varian Clinac iX on PRIMO

The first segment (*s*1) of the simulation focused on photon beam production and its interaction within the linac head. This commenced with the electron beam generated by the electron gun hitting the target material [31]. The beam parameters refer to the default values provided by the PRIMO simulation [23] such as an initial electron energy of 5.4 MeV, FWHM energy of 0 MeV, focal spot FWHM of 0 cm, and beam divergence of 0°.

The initial electron energy was varied at 5.2 MeV, 5.6 MeV, and 5.8 MeV to establish the optimal value. Around 4×10^8 particle histories were simulated to ensure statistical uncertainty within 1% in dose calculation. Although this led to longer simulation times due to the abundance of simulated particles, the trade-off was a reduction in dose uncertainties [32]. Variance reduction was employed through the splitting roulette technique, with the splitting region's size aligned with the field size set in segment-2. This technique enhances computational efficiency by reducing the particle weight, allowing for more precise tracking of the particles that have a more significant effect on the simulation results [33].

The second segment simulates the beam shaping, configuring the field size to approximately 10 cm × 10 cm, with the isocenter positioned at 0 cm along the XYZ direction and a source-to-surface distance (SSD) of 100 cm. The particle interactions in a phantom were simulated in segment-3. In this study, a water phantom measuring 40.2 × 40.2 × 40 cm^3 in size with a density of about 1 g/cm^3 was employed. The dose distribution was recorded at intervals of 0.2 cm along both the x and z directions. The phantom geometry is illustrated in Fig. 2.

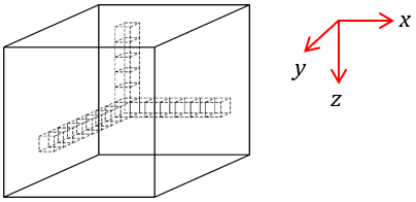


Fig. 2. Homogeneous Water Phantom Model in Segment 3

Subsequent to the simulation in segment-3, output data yielded one-dimensional dose distributions. The focus of analysis rested on two dosimetry parameters: PDD and dose profiles along the x-axis. The doses were computed using both DPM and PENELOPE, allowing for their accuracy to be assessed in comparison to measurements. The PDD obtained from the quotient between the dose at any depth to the quantity at reference depth as shown in (1).

$$PDD = \frac{D_d}{D_r} \times 100\% \quad (1)$$

where D_d is the dose at any point and D_r is the maximum dose in simulation. Additionally, the dose profiles were obtained at a depth of 10 cm. The dose normalized to 100% at the central beam axis. Computational efficiency was enhanced through the application of a splitting factor about 300 for DPM and 100 for PENELOPE. The cut-off energies for electron-positron and photon interactions were set at 0.2 MeV and 0.05 MeV, respectively.

2.2. Measurement

The measurements were performed using the Varian Clinac iX on 6 MV photon beam as the beam accelerator. This accelerator was installed at private hospital in Bandung, West Java. The relative measurements were conducted within a 3D water phantom (Blue Phantom II), manufactured by IBA Dosimetry, with tank dimensions of 67.5 × 64.5 × 56 cm^3 and a scanning volume of approximately 47.8 × 47.8 × 41 cm^3. The use of a large size water phantom is recommended as a reference medium for dosimetry, given its ability to capture scattered beams effectively [34], [35]. Two units of ionization chamber CC13, each with an active volume of 0.13 cm^3, were employed to measure the beam radiation output. These detectors were positioned in both the field (immersed in water) and reference (in air). The measurement were conducted at a 10 cm x 10 cm field size with a step size of 0.2 cm. The dose readings were acquired using a CCU electrometer. The measurement setup parameters were aligned with those of the MC simulation, including a gantry angle of 0°, collimator angle of 0°, and SSD of approximately 100 cm.

2.3. Linac Model Validation

The validation of the linac model encompassed a comparison of PDD and dose profiles between MC simulation and measurement [36]. The PDD compared to obtain the optimum initial electron energy parameters. The percentage difference between MC simulation and measurement was calculated using (2).

$$\Delta D\% = \left[\frac{(D_s - D_m)}{D_m}\right] \times 100\% \quad (2)$$

where D_s is the dose at any point in simulation and D_m is the dose at measurement point. The AAPM TG -105 protocol proposed the tolerance range of < 2% difference between the MC simulation and measurement [37], [38]. The dose profile was analyzed to verify the size and design accuracy of linac secondary collimator. The flatness and penumbra of the dose profile were also evaluated. The beam flatness was calculated using (3).

$$F = \left[\frac{D_{max} - D_{min}}{D_{max} + D_{min}}\right] \times 100\% \quad (3)$$

where D_{max} and D_{min} are the maximum and minimum dose point values at 80% beam width. The flatness value about < 3% require for linac with measurement in a water phantom at 10 cm depth and SSD 100 cm. The penumbra determine by calculating the distance between 20% and 80% dose on the beam profile [39]. The MC beam quality evaluated based on the tissue-phantom ratio ($TPR_{20,10}$) which known as beam quality index [31], [40]. The $TPR_{20,10}$ determined at reference field size with 100 cm SSD using (4).

$$TPR_{20,10} = \left[PDD_{20,10} \times 1.2661\right] - 0.0595 \quad (4)$$

where $PDD_{20,10}$ is the ratio of percentage depth dose at depth of 20 cm and 10 cm.

3. RESULTS AND DISCUSSION

The Monte Carlo simulation effectively generated depth dose and dose profile data of 6 MV photon beam for dosimetry analysis. The PDD and dose profile based on DPM algorithm for various initial electron energies are shown in Fig. 3.

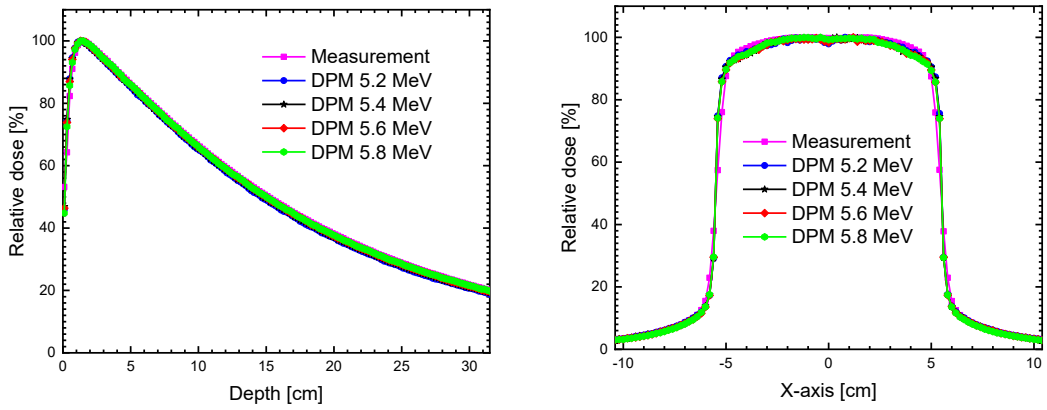


Fig. 3. Dosimetry Comparison for various initial electron energy with FS 10 cm × 10 cm and SSD 100 cm (a) PDD, (b) Dose profile

As expected, the depth dose exhibited an increase in the build-up region, attributed to electron and positron contamination effects. The maximum dose was observed at 1.3 cm depth for initial electron energy of 5.2 MeV, 5.4 MeV, 5.6 MeV, while it shifted to 1.5 cm depth for the energy of 5.8 MeV. Beyond this peak, the depth dose continuously decreased as it traversed the phantom. The dose differences relative to the measurement were 3.26%, 3.58%, 1.43%, and 0.98% for each energy, respectively. Consequently, the energy of 5.8 MeV was identified as the optimal initial electron energy due to its minimal PDD difference. The dose profile (Fig. 4b) exhibited a FWHM width of 11 cm, with mean difference of 3.02%, 3.19%, 3.85% and 4.63% for various energy configurations. The dose was flattened within the 80% beam width area, followed by a gradual decrease near the field size edge [41].

The subtle difference in the dose profile suggested a congruence between the size and design of the secondary collimator's in the PRIMO linac model with the actual Varian Clinac iX. The beam profile's flatness and penumbra at the optimum energy of 5.8 MeV were about 2.94% and 0.46 cm, respectively. Previous studies about Varian Clinac on 6 MV photon beam by Dirgayussa *et al.* [10], Mohammed *et al.* [42], Assalmi *et al.* [43] and Sarin *et al.* [44] stated that initial electron energy of 6.4 MeV, 5.6 MeV and 5.95 MeV, showing a good agreement with the measurement. Comparisons with prior studies indicated some divergence in optimal energies, likely influenced by disparities in linac geometry, transport parameters, and simulation engines. The MC simulation's third segment featured a comparison between DPM and PENELOPE engines at the optimal initial electron energy of 5.8 MeV. The result shows differences in simulation time and uncertainty for both engines, as presented in Table 1.

Table 1. Time simulation and uncertainty for DPM and PENELOPE in segment-3

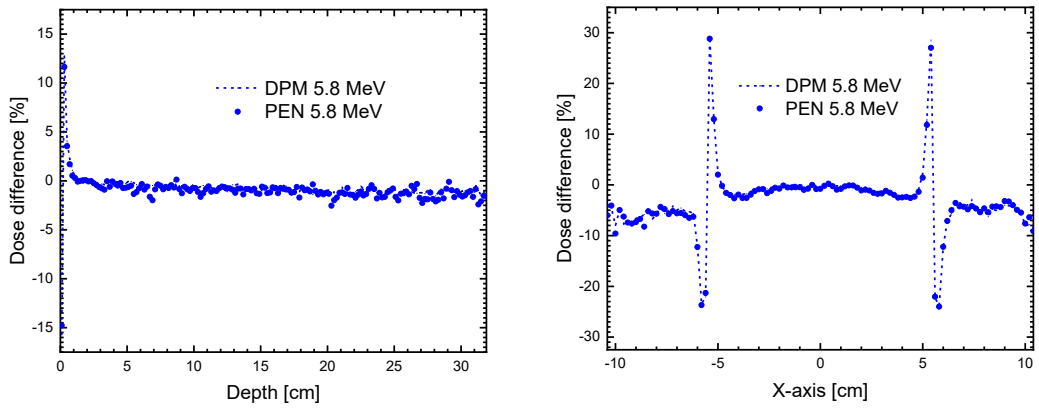
Simulation Engine	Time (s)	Uncertainty (%)
DPM	6514	0.59
PENELOPE	22610	0.80

The DPM's computational efficiency stood out, completing dose calculations around 3.5 times faster than PENELOPE while yielding a smaller statistical uncertainty of approximately 0.59%. The enhanced performance of DPM results from streamlining the particle transport algorithm and simplifying the underlying physics models. This result is in accordance with past study by Lopez *et al.* [23], where DPM was about 7 times faster than PENELOPE for VMAT plans simulation. The beam quality index for MC simulation and measurement shown in Table 2 were almost similar, with DPM provides a closer value to the measurement.

Table 2. Beam quality index calculation for measurement, DPM and PENELOPE

Method	$TPR_{20,10}$	Difference (%)
Measurement	0.66644	-
DPM	0.66358	0.42
PENELOPE	0.66255	0.58

The Fig. 4 showcased the dose differences relative to measurement for an initial electron energy of 5.8 MeV, with a field size of 10 cm × 10 cm and SSD of 100 cm, using both DPM and PENELOPE engines.



(a) (b)

Fig. 4. Dose difference relative to the measurement for initial electron energy 5.8 MeV, FS 10 cm × 10 cm and SSD 100 cm using DPM and PENELOPE (a) PDD and (b) Dose profile.

For PENELOPE, the mean differences in PDD and dose profile were 1.19% and 4.78%, respectively. The DPM calculation displayed slightly more accurate results to the measurement when compared with PENELOPE. This finding aligns with Rodriguez *et al.* [24] study, which highlighted DPM's efficiency with a dose difference of less than 1% compared to PENELOPE. These findings collectively underscore the efficiency and accuracy of DPM in dose calculation compared to PENELOPE, establishing a valuable contribution to the field. Nonetheless, in specific instances, PENELOPE could be the preferred option because the Klein-Nishina and Møller differential cross section utilized in DPM are only suited for low atomic numbers materials and high energies [24].

4. CONCLUSION

In this study, a comprehensive investigation into dose calculation accuracy using Monte Carlo (MC) simulations for Varian Clinac iX on 6 MV photon beam was conducted. The research aimed to determine the optimal initial electron energy, validate the linac model, and compare the efficiency of different simulation engines. The outcomes of the study established the optimum initial electron energy to be 5.8 MeV, as ascertained through Dose Planning Method (DPM), with an impressive PDD difference of about 0.98%. The evaluation of the linac secondary collimator design demonstrated its accuracy, revealing minimal deviations—less than 5%—between MC simulation and measurement for all considered energy variations. Comparing DPM and PENELOPE engines, the DPM exhibited a clear advantage. It not only enabled faster dose calculations in a water phantom but also demonstrated a greater precision in dose calculation consistent with

the measurement. This underscores DPM's efficiency and potential for enhancing dose calculation accuracy in radiotherapy.

To further enhance the understanding and application of these findings, future research avenues could explore varying other critical beam parameters, such as Full Width at Half Maximum (FWHM) energy, focal spot FWHM, and beam divergence. The diverse dataset generated in this study, including phase-space files, holds potential for contributing to subsequent investigations in the field. In summation, this study not only identified the optimal electron energy and validated the linac model but also illuminated the comparative benefits of simulation engines. The results generated contribute to refining the accuracy of dose calculations in radiotherapy, with implications for improving treatment planning and patient outcomes.

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