Monte Carlo simulation and experimental evaluation of dose distributions produced by a 6 MV medical linear accelerator.

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Abstract. The purpose of this work was to quantify the differences in dose distributions computed by Monte Carlo simulations against experimental measurements from an Elekta Synergy™ linear accelerator. The study was done using PRIMO, a PENELOPE-based Monte Carlo code. The dose calculation algorithm was compared under static field irradiations at 6 MV in a virtual water phantom for field sizes 10 × 10, 5 × 5 and 3 × 3 cm². Experimental depth doses and profiles were obtained using a pair of CC13 iba® ionization chambers at a SSD = 100 cm in an iba Blue phantom™. Simulations and experimental data were compared in terms of point by point differences. Gamma analysis was also used in order to evaluate dose-differences and distance to agreement (2%, 2mm respectively) of calculated and experimental dose distributions. The evaluation of depth dose distributions indicated that differences increased by decreasing the field size. In all cases the mean dose difference was below 1%. Lateral profiles differences were also below 1% for all field sizes. Gamma analysis results were in an agreement of 99% for almost all the dose distributions for the chosen criteria. The performed Monte Carlo simulation using PRIMO showed good agreement compared to experimental measurements for both, depth dose and dose profiles for the evaluated fields sizes, largely used in commonly radiotherapy treatments.

1. Introduction

Radiotherapy treatments requires a previous planning in order to generate beam shapes and dose distributions to achieve maximizing dose in the tumor region and at the same time minimize the dose in healthy tissues. These dose distributions are generated using dose calculations algorithms. The most common algorithms used in hospitals are based on analytical calculations [1], which provide reasonable accuracy requiring an acceptable amount of calculation time. Greater accuracy can be reached using Monte Carlo algorithms, especially in regions of large heterogeneities [2], but these algorithms require long simulation times and also involve difficulties preparing, executing and analyzing the results.

The aim of this work is to investigate the performance of Monte Carlo simulations for external radiotherapy in terms of dosimetric accuracy. This was done by benchmarking the results obtained in simulations versus experimental measurements obtained in a clinical environment.
2. Materials and methods

The treatment machine was an Elekta Synergy linear accelerator with 6 MV and 18 MV photon energies, as well as 4 MeV, 6 MeV, 8 MeV, 10 MeV, 12 MeV and 15 MeV electron energies. Square field sizes of 10 × 10, 5 × 5 and 3 × 3 cm$^2$ were both, simulated and measured.

2.1 Monte Carlo simulations

Calculations were computed using the PRIMO code [3], a computer software developed in 2013, that simulates clinical linear accelerators and estimates absorbed dose distributions in water phantoms and computerized tomographies (CT). It combines a graphical user interface and a computation engine based on the Monte Carlo code PENELOPE [4].

PRIMO allows to choose between different linear accelerators models implemented and two operations modes (electron or photon). In this work a 6 MV photon beam was used. The whole simulation is divided in three segments s1, s2, and s3. The s1 segment corresponds to the upper part of the linear accelerator (target, flattening filter, primary and secondary collimators and ionization chamber). The s2 segment simulates the movable collimators (jaws) and the multileaf collimator. The s3 segment is dedicated to dose estimation in a water phantom or CT. According to previous studies [5] an initial electron energy of 6.1 MeV, a FWHM (Full Width at Half Maximum) of 0.2 MeV for the beam energy and 3 mm FWHM for the focal spot size were chosen.

To increase the speed of the simulations, PRIMO has several variance-reduction techniques implemented. When simulating linear accelerator parts (s1 and s2), splitting roulette was used. When simulating the phantom segment (s3) a splitting factor of 300 was used.

2.2 Experimental measurements

Beam data acquisition was based on IAEA TRS-398 [6] recommendations. A pair of iba CC13 model were used as a reference and field ionization chambers. These detectors are of cylindrical geometry with active volume of 0.13 cm$^3$. The water phantom used in the measurements was a Blue Phantom from Scanditronix Wellhofer. Radiation beam incidence was perpendicular to the surface of water phantom. Fields were irradiated with 6 MV photon beam with a dose rate of 300 MU/min. The percent depth doses (PDD) and lateral dose profiles (crossplane) were acquired at depth of 5 cm and 100 cm source to surface distance (SSD). Dosimetric measurements and data were processed using OmniPro – Accept 6.6 software.

2.3 Data analysis

Comparisons were made between Monte Carlo simulations and experimental measurements. We used relative dose differences, the gamma index and z values for each point of the simulated and experimental curves.

The curve difference is calculated for a position $r$ as:

$$\Delta d(r) = \frac{d_e(r) - d_s(r)}{d_{e\text{max}}(r)} \times 100$$

where $d_e(r)$ and $d_s(r)$ are doses at the position $r$ of the experimental and simulated curves, respectively, and $d_{e\text{max}}(r)$ is the maximum dose of the experimental curve.

We also used the gamma index defined by Low et al. for two dose distributions [7] as follows,

$$\Gamma(r, r') = \frac{|r - r'|^2}{DTA^2} + \frac{(D(r) - D(r'))^2}{\Delta D^2}$$

$$\gamma(r) = \min \{ \Gamma(r, r') \} \forall r'$$

where $|r - r'|$ represents the distance between the points of experimental and simulated curves. DTA is the distance to agreement value, and $\Delta D$ is the dose tolerance value. In this work we used a DTA of 2 mm and a $\Delta D$ of 2% according to the study of Ahmad et al. [8].

Finally, we computed a statistical histogram defined by Ahmad et al. [8] as follows,

$$z = \frac{d_e(r) - d_s(r)}{d_s(r)} \cdot \frac{1}{\sigma_{tot}}$$
where $\sigma_{\text{Tot}}$ corresponds to the standard deviation of the whole distribution, for all the points of the experimental and simulated curves, which acts as a normalization factor.

3. Results and discussion

3.1. Monte Carlo simulations and experimental measurements

Simulated and experimental PDDs and dose profile curves are plotted in Fig. 1 for a 6 MV photon

![Figure 1](image-url)

**Figure 1.** 6 MV PDD curves (left) and lateral profiles at 5 cm depth (right) for field sizes of $10 \times 10$ cm$^2$, $5 \times 5$ cm$^2$ and $3 \times 3$ cm$^2$ at SSD = 100 cm. Monte Carlo data is plotted as points, while experimental measurements are plotted as a continuous line. Percent differences between the simulation and measurements are plotted in the lower panels.
Figure 2. Statistical histograms of dose differences calculated for dose distributions. The first row presents the results for PDD curves, while the second one for dose profiles.

| Field size | Averaged $\Delta d$ [%] | $\Gamma$ agreement [%] | $Z$ values [%] |
|------------|-------------------------|------------------------|----------------|
|            | PDD Profile              | PDD Profile            | PDD Profile    |
| $10 \times 10$ cm$^2$ | $0.17 \pm 0.40$ | $0.86 \pm 1.46$ | $0.61 \pm 1.0$ | $0.93 \pm 1.0$ |
| $5 \times 5$ cm$^2$ | $-0.70 \pm 0.71$ | $-0.18 \pm 1.4$ | $99.46$ | $99.71$ | $-0.96 \pm 1.0$ | $-0.73 \pm 1.0$ |
| $3 \times 3$ cm$^2$ | $0.66 \pm 0.61$ | $-0.58 \pm 2.0$ | $99.73$ | $100$ | $1.0 \pm 1.0$ | $-1.37 \pm 1.0$ |

beam for field sizes of $10 \times 10$, $5 \times 5$ and $3 \times 3$ cm$^2$ at 100 cm SSD. Each curve is normalized to dose at $d_{max}$ and differences between simulation and measurements are shown below each subplot. Table 1 shows the averaged $\Delta d$ for the PDDs curves. It can be observed that the mean value of dose differences is below 1 % in all cases. The negative sing in the $5 \times 5$ cm$^2$ field size indicates that doses calculated by Monte Carlo simulations overestimates the measured curve. Nevertheless, near the phantom surface, at the so called build-up region (before the maximum ionization dose), dose differences are of 4.04, 6.81 and 6.86 % for the field sizes of $10 \times 10$, $5 \times 5$ and $3 \times 3$ cm$^2$, respectively. This experimental overestimation is expected for cylindrical ionization chambers. Even the CC13 model (used in this work), with a considerable small cavity volume of 0.13 cm$^3$, overestimates the dose measurements at the build-up region, where a high gradient dose exists. This comes from the fact that at this section of the PDD curve, proximate to the surface of the water phantom and before $d_{max}$, an absence of charged particle equilibrium (CPE) exists [9]. Fig. 1 (right) shows lateral dose profiles at 5 cm depth. Evaluation of dose differences indicate that experimental and simulated curves varies in no more than $\pm 7$ % for the $10 \times 10$ and $5 \times 5$ cm$^2$ field sizes. In the case of the $3 \times 3$ cm$^2$ field size, the differences were about of $\pm 9$ %. Nevertheless, the difference mean value in all cases was below 1 %. The maximum differences for all the field sizes, are presented at the penumbra region, where a high gradient dose exists because of the rapid decrease at the edges of the radiation beam. Besides, for smalls fields few points were evaluated, thus increasing dose differences in the high dose gradient regions.
Table 1 shows the results of Gamma analysis (2% / 2 mm) applied to the PDD curves and lateral dose profiles. The agreement is very good for almost all the field sizes. In fact, Monte Carlo simulations and experimental measurements matched better than 99 %, except for the lateral dose profile of 5 × 5 cm$^2$ for which an agreement of just 91.58 % was achieved (Fig.1). An interesting point regarding these results is the effect of the dose voxel size. In order to simulate the *iba* Blue Phantom described in section 2, a virtual phantom of $48 \times 48 \times 30$ cm$^3$ (the actual size of the physical phantom) was generated in PRIMO. We defined a voxel size of 2 mm and the calculation time required for the entire simulation was about of 410 hours (~17 days) in an ordinary computer (3.4 GHz processor, 8 GB of RAM). Better results can be achieved if the voxel size is decreased, thus increasing voxel resolution, but increasing calculations times.

Results of $z$-values are shown in Table 1. In Fig. 2 we can observe that the histogram of differences for the PDD curves are close to a Gaussian distribution. However, only the $10 \times 10$ cm$^2$ dose profile seems to fit a Gaussian distribution. Moreover, the results of Table 1 indicate systematic differences in the simulated and measured curves. The maximum difference in this analysis is in the $3 \times 3$ cm$^2$ lateral dose profile. The statistical histograms visually indicate the distributions of dose differences spanning all voxels. If simulated and experimental curves are identical, we can expect a histogram perfectly fitted to a Gaussian distribution with the same standard deviation.

4. Conclusions

In this study a 6 MV Elekta linear accelerator was simulated using a PENELOPE based Monte Carlo code. Field sizes of $10 \times 10$, $5 \times 5$ and $3 \times 3$ cm$^2$ were evaluated for PDD curves and lateral doses profiles. Grater than 1 % dose differences at the build-up region were expected according to the lack of CPE. A good agreement of 99 % between Monte Carlo and experimental dose distributions was observed passing the 2 mm, 2 % criteria of the Gamma test, except for the $5 \times 5$ cm$^2$ lateral dose profile.

This first study demonstrates that PRIMO can be used in order to simulate radiation beams. By adjusting the initial parameters, more complicated radiation beams can be simulated in order to be used as a dosimetric tool for quality assurance.

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