DEPTH DOSE DISTRIBUTIONS OF THERAPEUTIC ELECTRON BEAM FROM VARIAN LINAC: MONTE CARLO STUDY AND EXPERIMENTAL MEASUREMENTS

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Abstract— **Linac-based electron beam therapies are used for the treatment of superficial cancer tumors. Central axis depth dose distributions for 6, 9, 12, and 15 MeV nominal electron energies delivered by Varian Clinac iX with 10 × 10 cm^2 applicator were computed by using the MCNPX (V. 2.6.0) Monte Carlo code. Percent Depth Dose (PDD) distributions computed by MC simulations were validated through comparison with the corresponding measured data. Discrepancies between MCNP and experimental data were found within 1.62% and 1.31 mm in the therapeutic range (90 to 80% of maximum dose values) of electron beam and these are within the recommended standard (±2%) used in the dose calculation. However, notable variations were found beneath the depth of 50% dose, especially towards the bremsstrahlung tail region which is not normally considered in the treatment planning system. The deviations at the high-dose gradient region might be due to scattering foils and collimator jaws during MC modelling, resulting in the lower production of bremsstrahlung photons. As the MC computed data were in good agreement with experimental values except for the high-dose gradient region, the developed Monte Carlo program can be used in the various dosimetric study of the therapeutic electron beam in a homogenous and inhomogeneous media as well as to investigate the contaminations of photons and neutrons during the treatment.**

Keywords— **Electron beam, Dose distributions, Monte Carlo simulation, MCNPX, Varian linac.**

I. INTRODUCTION

Electron beam therapy is an important modality for the treatment of superficial tumors (less than 5 cm deep). It is extensively used for chest wall irradiation of breast cancer, skin and lip cancers, head and neck cancers, etc. [1]. Modern external electron beam therapy is carried out using a medical linear accelerator. Treatment planning in radiotherapy is an important procedure to evaluate the dose in a patient before performing the actual treatment. Central axis depth dose distributions of the electron beam are evaluated during the treatment planning of superficial cancerous cells. In early methods, central axis dose distributions of electron beams were carried out experimentally using phantom which was time-consuming. Later on, various algorithms such as Pencil Beam, Pencil Beam Redefinition, Collapsed Cone Convolution algorithms, etc., were developed to calculate dose distributions theoretically and thus improved efficiency [2]. However, in radiation therapy, it is recommended that the accuracy of dose delivery to cancer cells should be within ±5% [3]. To obtain it, the accuracy in dose calculation must not exceed ±2%. But the complexity of electron-tissue interactions makes it very difficult to obtain such accuracy by using conventional treatment planning algorithms. Currently, the Monte Carlo algorithm is used as the most accurate method for calculating dose distribution in radiotherapy. The MC algorithm can reduce the uncertainty in the dose calculation within recommended values as it takes into account the multiple scattering and the creation of secondary particles or delta rays in electron dosimetry [4].

Various studies have been performed on central axis dose distributions of the electron beam in water phantom by Monte Carlo simulations and experimental techniques. Toossi *et al.* [5] simulated Siemens Primus linac by using MCNPX (version 2.4.0) MC code for 8, 12, and 14 MeV electron beams with various electron applicators (10 × 10 cm^2, 15 × 15 cm^2, 25 × 25 cm^2). The PDD data from their simulation were in good agreement with experimental ones. The maximum discrepancy between the simulated and measured values of R_{50} was 1.3 mm at 10 × 10 cm^2 applicator. Nedaie *et al.* [6] performed an MC simulation of ELEKTA Precise linac using MCNP4C code to investigate the effect of various components of linac head and efficacy of MC algorithm in producing dosimetric data. They used 8 & 15 MeV electron beams and a 10 × 10 cm^2 treatment field to get percent depth dose (PDD) data and beam profiles. A p-type diode detector was used in getting experimental data. The discrepancy between the simulated and experimental PDD was within 2% and they concluded that to get better results theoretically, all the main components of the linac head must be added in simulation geometry. Lalić *et al.* [7] investigated central axis depth dose distribution in water for 6, 9, and 12 MeV electron beams from a Varian 2100C medical linac. They utilized FOTELP MC code in their simulation. Due to the unavailability of a high-speed computer, they made some simplifications in the geometry of the linac head and used a lower number of electron histories. As a result, their simulation results have significantly differed from their experimental values. Aziz *et al.* [2] performed both the experimental and theoretical

calculations of depth dose distribution along the beam central axis in a homogenous 3-D water phantom for a 9 MeV electron beam from a Seimens Primus medical linac. They used both BEAMnrc and DOSXYZnrc source code in the EGSnrc MC package. Simulation results were in good agreement with the measured data obtained by an ion chamber and the maximum discrepancy was less than 2%.

This work aims to simulate the treatment head of Varian linac by using MCNPX code for 6, 9, 12, and 15 MeV electron beams and to compare the Monte Carlo calculated central axis depth dose data with the corresponding experimental values obtained by the PPC40 plane-parallel ion chamber.

II. METHODOLOGY

Monte Carlo Simulation

The treatment head of an electron mode Varian CLINAC (model IX) was simulated by using the MCNPX (version 2.6.0) Monte Carlo code [8]. The details of Varian linac head description are given elsewhere [9]. The Varian IX CLINAC has a two-photon mode (6, 10 MV) and several electron modes. Among them four electron beam energies 6, 9, 12, and 15 MeV were selected for simulation purposes. The major components of the accelerator head such as electron scattering foils (the primary foil made of tantalum, and the secondary foil made of aluminum), a 6.77 cm long primary conical collimator (made of tungsten), secondary collimator pairs with a thickness of 7.77 cm (made of tungsten) and 10 $\times 10$ cm^2 electron applicator were simulated in this study. The materials compositions, shapes, and dimensions of these components were collected from the technical drawing manual of linac provided by the Varian medical system [10]. Moreover, a standard cubic water phantom (40 $\times$ 40 $\times$ 40 cm^3) was included as a part of head components. A monoenergetic and monodirectional beam with a radius of 1 mm was used as a primary electron source. Figure 1 shows the geometry of accelerator head components considered in this study.

To compute the absorbed dose values, a series of cylinders, with 1 mm height and 1 cm radius, were modeled on the central beam axis of the water phantom. The *F8 tally was used to calculate the depth dose distribution of the electron beam. PDD values were obtained by normalizing the MC calculated dose values to the maximum dose at arbitrary depth on the beam central axis and multiplied by 100. Several input files were run with at least 3E8 source particles and the average statistical uncertainty was within 3%. The whole simulation work was carried out with an Intel Core i9 processor desktop computer.

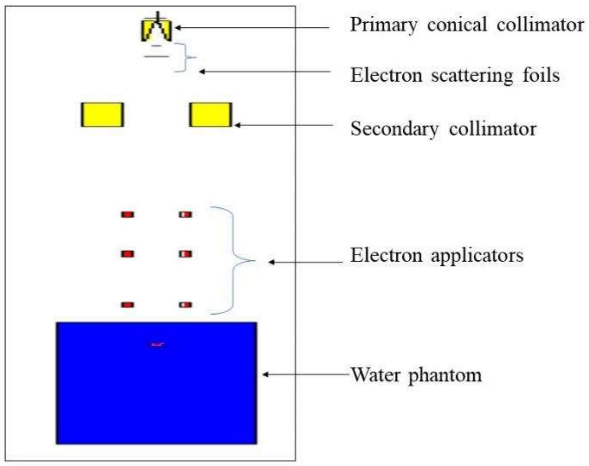

Figure 1. Simulated accelerator head components of electron mode Varian CLINAC

Experimental Techniques

Dose measurements were carried out in an IBA blue phantom 3D water phantom by using a PPC40 plane-parallel ion chamber (0.6 cc) at Nuclear Medical Physics Institute (NMPI), Savar, Bangladesh. The water phantom was set at 100 cm SSD (source to surface distance) and projected a 10×10 cm^2 treatment field vertically at the phantom surface. After readjusting the effective point of ion chamber to 100 cm SCD, the ion chamber charge readings were taken with a 1 mm step from the phantom surface towards its bottom along the beam central axis until a constant value was reached. These charge readings were then converted to dose values by using the appropriate stopping power ratio, water to air, according to the TRS-398 code of practice [11]. The experimental setup for depth dose determination is shown in Figure 2.

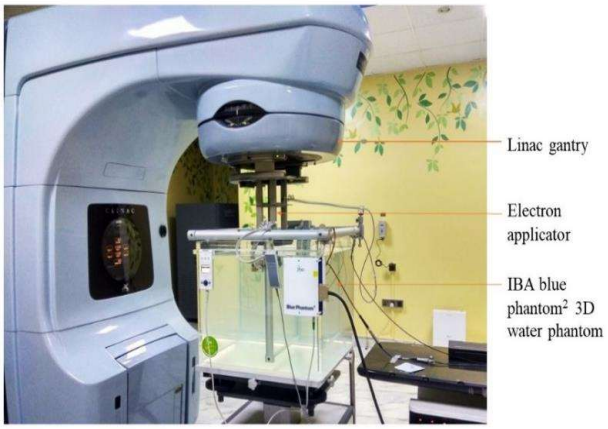

Figure 2. Experimental set-up for obtaining electron beam depth-dose distributions

III. RESULTS

The calculated and measured central axis depth dose distributions for 6, 9, 12, and 15 MeV electron beams with 10×10 cm^2 applicator are presented in Figure 3. Dose values at different depths were normalized to the maximum dose. From this figure, it was seen that the MC calculated absorbed dose values were in excellent agreement with the measured ones up to 50% of the maximum dose in 6, 12 and 15 MeV energies.

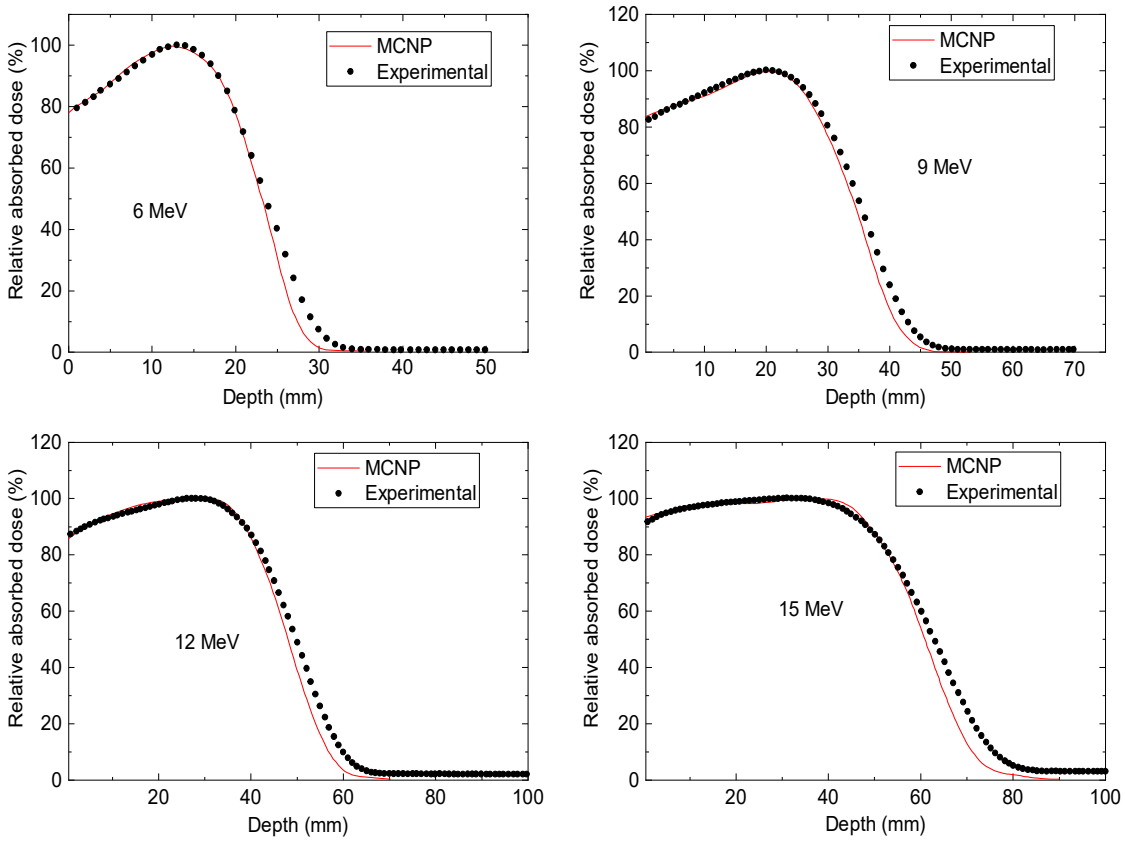

Figure 3. PDD distributions of 6, 9, 12, and 15 MeV electron beams for 10×10 cm^2 applicator

A point-to-point comparison between MCNP and measured doses up to D_{max} shows the differences within 1.27%, 1.44%, 1.62%, and 1.27% for 6, 9, 12, and 15 MeV nominal electron beams respectively. Dose differences at D_{max} position were found to be 0.53 mm, 0.29 mm, 0.85 mm, and 1.31 mm for these energies respectively. The maximum deviation in dose determination at the R_{50} position was 0.19 mm which was related to 12 MeV energy. Figure 3 also shows that the dose discrepancies were more pronounced in the high dose-gradient region. It could be due to the lower production of bremsstrahlung photons during MC simulation. As we know the most bremsstrahlung photons are mainly produced in the scattering foil and the collimator jaws, so the small differences in the declaration of these thicknesses can create an inappropriate prediction of the contaminated photons. However, these regions are not important in treatment planning with the electron beam. Most importantly, the discrepancy in the electron beams therapeutic range (90% - 80% of maximum dose values) was within 2% which is the acceptable standard for dose evaluation. From figure 3, we also evaluated some important electron depth dose parameters such as mean energy $\bar{E}_0$ on the phantom surface, depth of maximum dose (Z_{max}), depth of 90% dose level (R_{90}), and electron beam quality index (R_{50}). Among them $\bar{E}_0$ was calculated using the following expression [12]:

$$\bar{E}_0 = 2.33\, R_{50} \quad (1)$$

These parameters are shown in Table.

From this table, it was observed that the maximum difference of R_{90} values between these two sets of data was 1.9 mm which was related to 12 MeV energy. The maximum discrepancy of $\bar{E}_0$ and Z_{max} was 0.44 MeV and 4 mm corresponding to 9 and 15 MeV respectively.

Table. Typical depth dose parameters of clinical electron beams

Depth dose parameters	Nominal Electron Energy							
	6 MeV		9 MeV		12 MeV		15 MeV	
	MCNP	Measurement	MCNP	Measurement	MCNP	Measurement	MCNP	Measurement
R_{90}	17.99 mm	17.98 mm	26.6 mm	27.48 mm	38.59 mm	38.59 mm	48.59 mm	48.12 mm
R_{50}	23.2 mm	23.7 mm	34.7 mm	35.6 mm	47.9 mm	49.8 mm	61.1 mm	62.8 mm
Z_{max}	12 mm	13 mm	20 mm	20 mm	25 mm	28 mm	35 mm	31 mm
$\bar{E}_0$	5.41 MeV	5.52 MeV	8.08 MeV	8.29 MeV	11.16 MeV	11.60 MeV	14.24 MeV	14.63 MeV

IV. CONCLUSION

This study presented the MCNP simulation of electron mode of a Varian CLINAC for 6, 9, 12, and 15 MeV energies in a homogenous medium and it was validated by comparing the computed dose distributions with experimentally measured values. From the analysis of our findings, we observed that the discrepancies were within 1.62% and 1.31 mm in the treatment regions of the electron beam and it is fulfilled the criteria (2%/2mm) which is mainly used in the commissioning of Monte Carlo based dose calculation [13]. However, the maximum discrepancy was observed in the high-dose region especially in the tail part of the PDD curve. This might be due to scattering foils and collimator jaws during MC modelling, resulting in the lower production of bremsstrahlung photons at higher depth. But in fact, the target cells are not generally located in this region so that the large discrepancy in dose evaluation in this part does not affect the precession of dose delivery to the targeted tumors but a contribution to the normal cell. Thus, the good agreement between the calculated and measured results encourages the use of MCNP Monte Carlo code as a reliable dose predictor where experimental measurements may not be easily feasible. Moreover, the developed MC program could also be used to study the dose distribution in a heterogeneous medium and to investigate the contamination of other particles during the treatment by a high-energy electron beam.

ACKNOWLEDGEMENT

The authors would like to express their sincere thanks to the staffs and scientists of Secondary Standard Dosimetry Laboratory of Bangladesh Atomic Energy Commission for their strong support in performing this research work. This work was supported by ADP project “Establishment of Calibration and Quality Control Facilities for Radiotherapy, Diagnostic Radiology and Neutron” of Ministry of Science and Technology, Bangladesh and Department of Physics, Faculty of Science, University of Rajshahi, Bangladesh.

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