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Beam Quality Indices for Therapeutic Electron and Photon Beams: A Monte Carlo Study

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Abstract
Background: *Electron and photon beam therapies are important modalities for the treatment of cancer patients. Current dosimetry protocols (TRS-398, AAPM, IPEM, etc.) recommend the tissue phantom ratio $TPR_{20,10}$ and R_{50} as beam quality specifier for therapeutic photon and electron beams respectively. This work presents the Monte Carlo simulated and experimental values of $TPR_{20,10}$ for 6 and 10 MV photon beam and R_{50} values for 12 and 15 MeV clinical electron beam.*
Materials and methods: *Varian 2300 CD Cinac and two ion chambers (FC65-G and Exradin A10) were modelled by using the MCNPX (V. 2.6.0) Monte Carlo code. Experimental measurements were also carried out for the same chambers using the PTW 3-D water phantom.*
Results: *Good agreement was obtained between the calculated and measured values. The maximum discrepancy of $TPR_{20,10}$ and R_{50} values between two set of data were 3.12% for 6 MV photon beam and 1.68% for 15 MeV respectively. The variation between simulated and experimental central axis depth-dose data for electron beam upto D_{max} were within 1.02687% and 1.54028% for 12 and 15 MeV energies respectively.*
Conclusion*: As the variations of these data set is small, the developed Monte Carlo program can be used in various dosimetric study of photon and electron beams in homogenous media.*
Key word: *Beam quality index; Medical linac; Monte Carlo simulation; MCNP; TRS-398.*

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I. Introduction

Modern radiotherapy treatment modality mainly employs high-energy electron linear accelerators for the irradiation of cancer cells. Superficial cancerous cells are often treated with electron mode Linac[1]. However, deep-seated tumors are often treated with X-rays produced by the bremsstrahlung interaction of electrons with a high-Z target material[2]. Various dosimetric quantities like wall correction factor, stopping power ratio, central electrode correction factor, the quality conversion factor k_Q, etc., depend upon the quality of therapeutic electron or photon beams[3]. Thus, the beam quality must be specified for clinical electron and photon beams from medical Linac. Most recent dosimetry protocols based on absorbed dose to water calibration of ion chambers use the half-value depth R_{50} in water as the beam quality index for clinical electron beam and the tissue phantom ratio, $TPR_{20,10}$, as the high energy photon beam quality specifier (IPEM, IAEA TRS 398, etc.)[4,5]. The electron beam quality index R_{50} is defined as the depth in water (in cm) at which the electron beam depth dose reduces to 50% of its highest value, measured with a constant SSD (source to surface distance) of 100 cm and a reference field size at the phantom surface. The photon beam quality index $TPR_{20,10}$, on the other hand, is defined as the ratio of absorbed dosage to water at depths of 20 and 10 cm in a water phantom obtained with a constant source to chamber distance (SCD) of 100 cm and a field size of 10×10 cm^2 at the detector position. The beam quality index $TPR_{20,10}$ is notable for its independence from electron contamination in the incident beam[6,7,8].

Monte Carlo algorithms have been extensively utilized in radiotherapy for accurate modeling of linear accelerator and estimating dose distributions for treatment planning. MC simulation is a well-established technique for benchmarking photon and electron dose estimations in therapeutic treatment. Several researchers have been used the Monte Carlo method to determine the electron beam quality specifier R_{50} as well as the tissue phantom ratio ($TPR_{20,10}$) by modeling therapeutic linac machine. In a paper by Fonseca et al.[9] - a Monte Carlo modeling expert group (MCMEG) presented the values of $TPR_{20,10}$ for 6 MV photon beam by using different MC codes like MCNP6, MCNPX, PENELOPE, and EGSnrc, etc. In their comprehensive study, the photon beam quality index was computed by using several published photon spectra. $TPR_{20,10}$&$PDD_{20,10}$ values were also measured for Varian 2300 CD linac by using the PTW30013 ion chamber. Wulff et al.[10] determined the values of $TPR_{20,10}$ for 4, 6, 10, 15, 18 MV photon beams by using a simplified MC model developed in Beamnrc Monte Carlo code. In addition, a complete linac machine- Siemens KD was also simulated for 6 and 18 MV photon beam to evaluate $TPR_{20,10}$ values. The electron beam quality index R_{50} for 8, 12, and 14 MeV clinical electron beams from Siemens Primus linac was performed by Toossi's group using MCNPX MC code[11]. The maximum discrepancy of R_{50} values between MCNP and measurement was 1.3 mm.

The current investigation aims to determine the values of quality indices R_{50} and $TPR_{20,10}$ for 12, 15 MeV electron and 6, 10 MV photon beams respectively by complete modeling of a Varian 2300 CD Linac using MCNPX (V.2.6.0) Monte Carlo code and to compare the simulated values with the experimental ones obtained by Exradin A10 and FC65-G ion chambers.

II. Material and Methods

Monte Carlo simulation

The accelerator treatment head of a Varian Linac 2300 CD was simulated by using the MCNPX (V. 2.6.0) Monte Carlo code [12]. Materials compositions and dimensions were according to the specifications depicted on the technical drawing manual of Varian medical system [13]. The modeling of geometry of this Linac was created by using the standard macrobody surfaces BOX, RCC, SPH, TRC, etc., of the MCNP geometry specifications parameters. Varian 2300 CD Linac usually operates in 6 and 10 MV nominal photon energies and 4, 6, 10, 12, 15, and 18 MeV nominal electron energies. Among them two electron energies 12 & 15 MeV and both photon energies were selected for the simulation purposes. The major components of photon mode Linac which were modeled are the bremsstrahlung target, the primary conical collimator, the flattening filter, and secondary collimators. On the other hand, the simulated components of electron mode Linac were the scattering foils, the primary collimator, secondary collimators, and the 10×10 cm^2 electron applicators. The primary source electrons were modeled as a monoenergetic beam of radius of 2 mm which was directed vertically towards the tungsten target to produce bremsstrahlung photons. Moreover, a water phantom of box shape with dimensions 30×30×30 cm^3, a FC65-G thimble type cylindrical ion chamber, and an Exradin A10 parallel ion chamber were also modeled to compute the dose distributions within water phantom. The technical specifications of these ion chambers

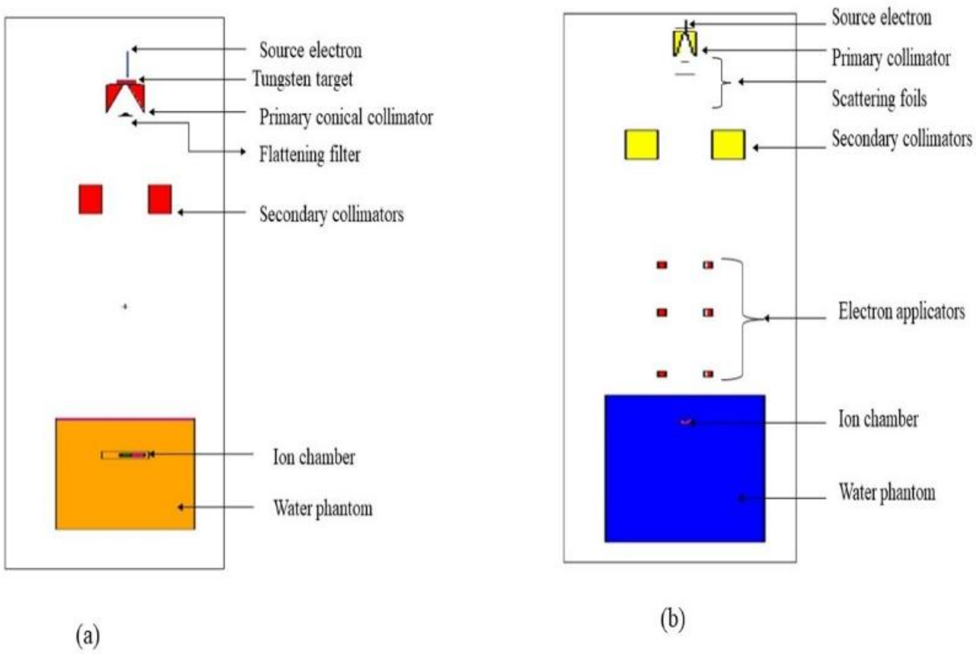

Figure 01: Simulated accelerator head components of Varian iX Linac (a) photon mode, (b) electron mode.

were collected from the various published articles and manufacturer's website. Figure 01 shows the simulated accelerator head components of both photon & electron mode including chambers. Figure 02 presents the simulated FC65-G chamber with components materials and dimensions. For $TPR_{20,10}$ estimation, the dose values in the sensitive volume of FC65-G detector at 10 cm and 20 cm depths in the water phantom were calculated by using the F4 photon tally for both 6 MV and 10 MV nominal photon beams in a SCD-type setup (field size 10×10 cm^2, SCD 100 cm).

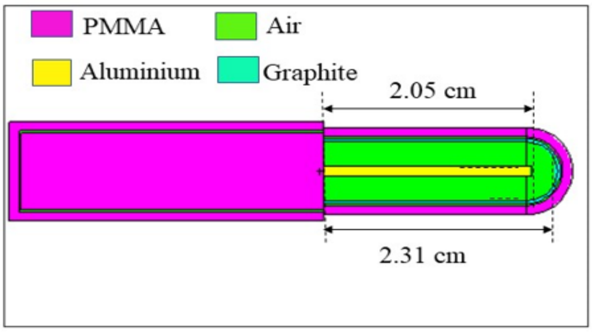

Figure 02: Simulated model of FC65-G thimble chamber.

However, the F4 tally calculates the average fluences in a cell, so that the additional FM4 photon tally multiplier was used to convert the fluences into the absorbed doses. The cut-off energy and physics cards for photons and electrons transport were kept at its default values.

For R_{50} estimation, the PDD distributions of electron beams were first obtained by computing the absorbed doses in the sensitive volume of Exradin A10 detector at different depths of water phantom with 1mm interval and normalizing these values by the maximum dose at arbitrary depth along the beam central axis. The F4 tally including FM4 tally multiplier was used to calculate the depth dose distribution of the electron beam. From the simulated PDD distributions, the R_{50} values were obtained by locating the depth at which the dose falls to 50% of its maximum value. A number of input files were run with NPS (number of source particles) at least 20×10^7 until the average statistical uncertainty in the tally cells were within 3%. The whole simulation works were carried out on an Intel Core i9-9900K CPU @ 3.60 GHz 32.0 GB RAM desktop computer.

Measurement techniques

Photon beam quality index $TPR_{20,10}$ was measured for 6 and 10 MV photon beams from Varian Linac 2300 CD at the National Cancer Research Institute & Hospital (NIRCH) in Dhaka, Bangladesh. Measurements were carried out using an IAEA standard cubic water phantom (30×30×30 cm^3) and a calibrated FC65-G Farmer type ionization chamber of 0.6 cm^3 sensitive volume. This ionization chamber was connected to a IBA dose-1 dosimeter. The water phantom was positioned below the Linac gantry and the ion chamber was immersed in the water phantom along the beam central axis at depths of 10 cm and 20 cm. The reference conditions for $TPR_{20,10}$ measurements were according to the *TRS-398* dosimetry protocol [14]. The $TPR_{20,10}$ was determined using the following equation:

$$TPR_{20,10} = \frac{Q_{20}}{Q_{10}}$$

where Q_{20} and Q_{10} are the ion chamber readings corrected for various influence quantities like pressure, temperature, and ion recombination, etc., at depths 10 and 20 cm, respectively.

Measurements of depth ion distributions for electron beam were performed in an IBA blue phantom[2] 3D water phantom using an Exradin A10 plane-parallel ion chamber with 0.6 cm^3 sensitive volume. The reference conditions were in accordance with the *TRS-398* dosimetry protocol. The ion chamber charge readings had been taken at 1 mm interval 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. The experimental set-up for depth dose determination is shown in Figure 03.

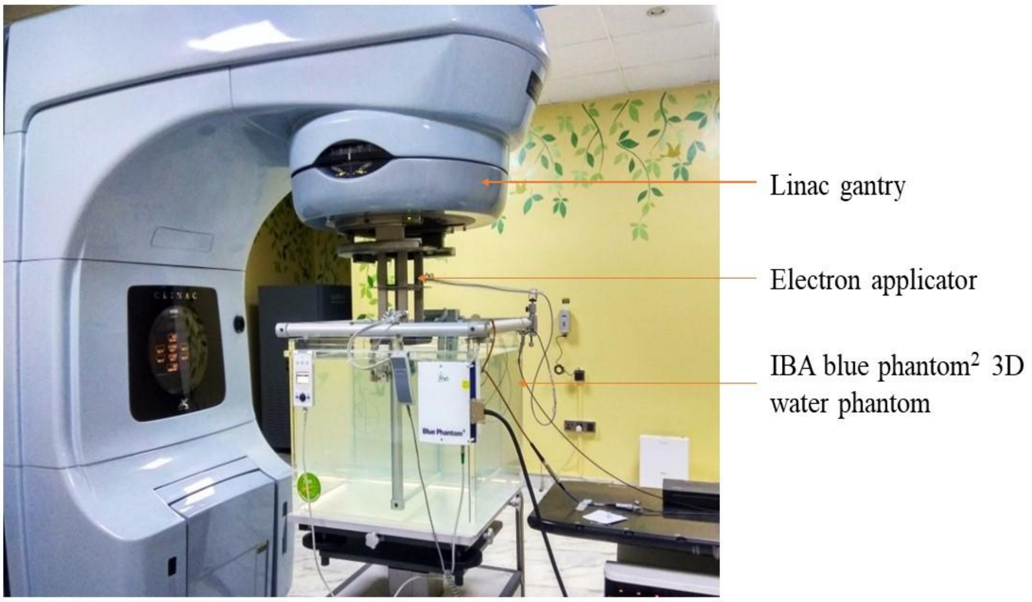

Figure 03: Experimental set-up for electron beam depth dose distributions.

III. Results and Discussions

Photon Beam Quality Index *TPR₂₀,₁₀*

Monte Carlo calculated values of beam quality index $TPR_{20,10}$for 6 and 10 MV photon beams were determined from the dose-energy spectra shown in Figure-1. The dose rate -energy spectra were plotted for both energies at 20 and 10 cm depths of water phantom using the OUTP file created after successful run of MCNPX input file. The values of dose rate for 6 MV photon beam at both depths were obtained and given as $\dot{D_{20}} = 0.405732 \pm 0.013000\ [Gy.hr^{-1}]$&$\dot{D_{10}} = 0.585939 \pm 0.103400\ [Gy.hr^{-1}]$. By taking the ratio of these two quantities, the tissue phantom ratio was obtained as 0.692 ± 0.182. The values of dose rate at 20 and 10 cm depth of water phantom for 10 MV photon beam were calculated as $\dot{D_{20}} = 1.67208\ \pm 0.09340\ [Gy.hr^{-1}]$ and $\dot{D_{10}} = 2.32234 \pm 0.06610\ [Gy.hr^{-1}]$. Dividing these dose values, the tissue phantom ratio $TPR_{20,10}$for 10 MV photon beamwas obtained as 0.720 ± 0.027. From the values of tissue phantom ratio, we observed that $TPR_{20,10}$increases with the increasing of incident photon energy. When we looked at dose values (alternatively charge values) at 20 cm depth (reference point of the modelled FC65-G chamber) for two photon modes, it was seen that the dose deposited by higher energy photon beam was greater than that of the low energy photon beam. The directly ionizing radiation produces in the outside of the cavity through photon interactions have higher energy so that it can penetrate a larger depth of water. Moreover, the production of bremsstrahlung photons in wall material by these radiations also interact with the chamber air molecules and central electrode material, consequently, increases the deposition of doses. On the other hand, the directly ionizing radiation produces from the low energy photon cannot penetrate a larger depth because they deposit their energy at a smaller depth (locally) so that the dose deposition at the desired depth was also small.

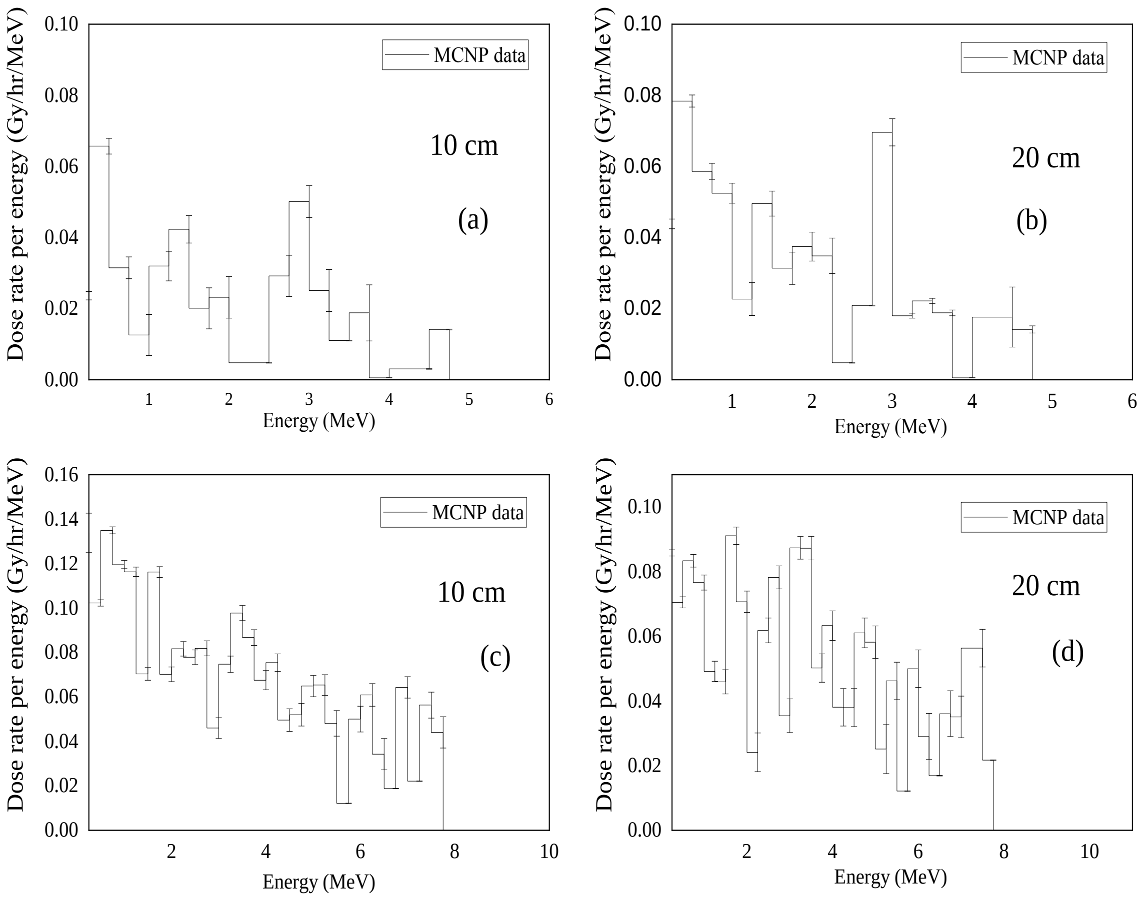

Figure 01: Dose-energy spectra for 6 MV (a & b) and 10 MV (c & d) photon beams at 20 and 10 cm depths of water phantom.

Electron Beam Quality Index R_{50}

Simulated and experimental relative abosrbed dose data for 12 and 15 MeV clinical electron beams are presented in Figure 02. These dose distributions were obtained for 10 × 10 cm^2 electron applicator by modelling an Exradin A10 plane-parallel ion chamber. Dose data at different depths were normalized to the maximum dose values. From these graphs, it was observed the relative dose increases upto a maximum value at a certain depth. Beyond this depth dose distribution falls down sharply. The point of maximum dose was close to the surface of water phantom for low energy particle. Electron beam of low energy deposits maximum kinetic energy in the surface region due to its multiple scattering characteristics. On the other hand, high energy electron beam penetrates and loses its kinetic energy over a region of higher depth because it suffers minimal scattering near the surface region of the phantom. A comparison between simulated and experimental dose data upto D_{max} shows the differences within 1.02687% and 1.54028% for 12 and 15 MeV nominal electron beams respectively. The maximum dose discrepancy at D_{max} was 5.00346 mm for 15 MeV beam. However, the discrepancy of absorbed dose was more pronounced in the high-dose gradient region. The constant dose distributions was also observed beyond the rapid dose fall down region which is called the bremsstrahlung tail. This constant region is responsible for the photon contamination of the incident electron beam with the scattering foils, secondary collimators, electron applicators, and air between accelerator window and the phantom surface. From these graphs, the beam quality index R_{50} for electron beam was obtained by locating the position of the depth at which the dose falls down to half of its maximum value. Table 01 shows the measured and calculated values of beam quality indices for both photon and electron beams.

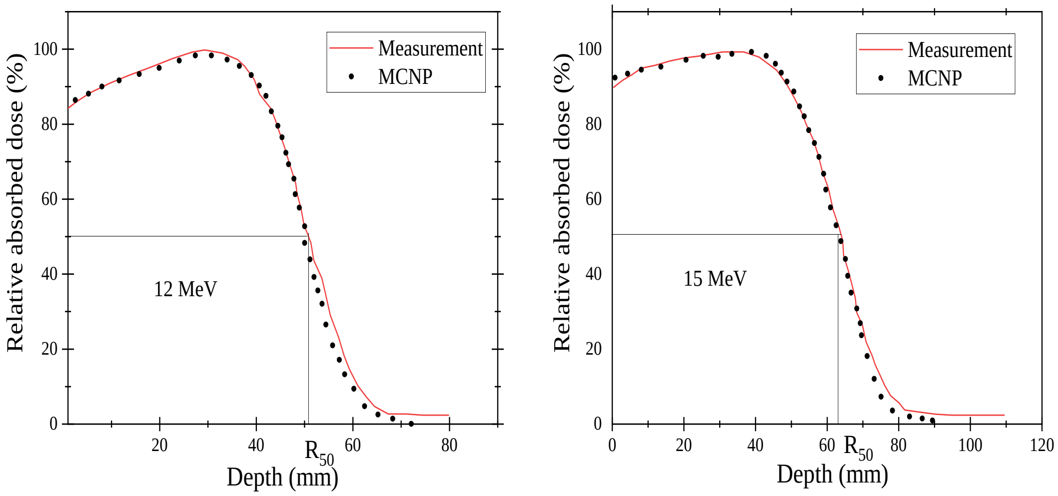

Figure 02: Depth-Dose distributions of 12 and 15 MeV therapeutic electron beam in water phantom.

Table 01: Beam quality indices for photon and electron beams

Energy (MeV)	Chamber	Beam quality indices			Uncertainty
6	FC65-G	$TPR_{20,10}$	Measurement	0.671	3.12 %
			MCNP	0.692	
10			Measurement	0.740	2.7 %
			MCNP	0.720	
12	Exradin A10	R_{50} (mm)	Measurement	50.63	1.18 %
			MCNP	49.45	
15			Measurement	63.72	1.68 %
			MCNP	62.04	

IV. Conclusion

This study was mainly focused on Monte Carlo simulation of Varian Clinac for two photon beam (6 and 10 MV) and two electron beam (12 and 15 MeV). Using the simulated linear accelerator as well as two modelled ion chamber FC65-G and Exradin A10, the beam quality indices $TPR_{20,10}$ and R_{50} were determined and compared with the experimental measurements performed in a 3-D water phantom. Good agreement between calculated and measured values of the tissue phantom ratio and R_{50}encourages the use of Monte Carlo calculated data in treatment planning as a reliable dose predictor as well as in dosimetry purposes where the experimental measurements are not easily feasible.

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