International Journal of Medical, Medicine and Health Sciences
ISSN: 2517-9969
Vol:12, No:4, 2018

A Dose Distribution Approach Using Monte Carlo Simulation in Dosimetric Accuracy Calculation for Treating the Lung Tumor

Md Abdullah Al Mashud, M. Tariquzzaman, M. Jahangir Alam, Tapan Kumar Godder, M. Mahbubur Rahman

***Abstract*—**This paper presents a Monte Carlo (MC) method-based dose distributions on lung tumor for 6 MV photon beam to improve the dosimetric accuracy for cancer treatment. The polystyrene which is tissue equivalent material to the lung tumor density is used in this research. In the empirical calculations, TRS-398 formalism of IAEA has been used, and the setup was made according to the ICRU recommendations. The research outcomes were compared with the state-of-the-art experimental results. From the experimental results, it is observed that the proposed based approach provides more accurate results and improves the accuracy than the existing approaches. The average %variation between measured and TPS simulated values was obtained 1.337±0.531, which shows a substantial improvement comparing with the state-of-the-art technology.

***Keywords*—**Lung tumor, Monte Carlo, polystyrene, elekta synergy, Monaco Planning System.

I. Introduction

EFFICIENT dose calculation is the most challenging task in the world for clinical medical physicist. The ultimate goals of the physicists are to deliver the maximum dose to the cancerous tissues and minimum dose to the normal tissues and risk organs and to save the normal tissue. This approach is engaged in variety of cancer treating research.

One of the most significant parameters in dosimetric analysis is Tissue Phantom Ratio (TPR) measurement [1]. Instead of using the available conventional expression, this paper chooses the most valuable 2-point measurement expressions and adaption of interpolation for that purpose. There are many advantages for 2-point measurement and adaption of interpolation such as reducing the computational time, improving the precision and monitoring the quality assurance (QA).

The two well-known patient setup techniques for delivering dose to the carcinoma patients are Source to Surface Distance (SSD) technique and Source to Axis Distance (SAD) or iso-centric technique [2]. The SSD technique is more convenient to assess the percent depth dose (PDD), but it is inconvenient for patients. Actually, human body is inhomogeneous and different organs are sited at different depth from the body surface. On the other hand, the SAD technique is very convenient for delivering dose to the patients. However, the SAD technique required high number of measurements.

Besides, several commercial treatment planning systems use various types of dose calculation algorithm [3]-[5]. The most useful algorithms are three dimensional convolution [6], [7], collapsed cone convolution (CCC) [8], anisotropic analytical algorithm (AAA) [9] and MC Methods [10]-[15]. For high calculation speed, the CCC method is widely applied in commercial treatment planning systems. However, the differences greater than 5% have found at the materials for several densities such as between lung and tissue [16], [17]. The 5% difference is deemed unacceptable for better dosimetry to deliver the dose to the patients [18], [19]. The dose calculation speed is comparatively slow in the MC method. However, it has better accuracy compared to the other algorithms [20]. MC based calculation focuses on the inhomogeneity correction [21]. Photon beam dose calculation which is formulated by using MC methods [22] has adapted the American Association of Physicist of Medicine. The MC base dose calculation been has suggested [23] for commercial treatment planning systems and for the air cavity measurement [24]. Dosimetric accuracy calculation and QA has been estimated using AAA algorithm [25], here, the difference is found to be less than 5% but very greater than 2%. The accuracy is recommended less than 2% for better treatment [26]. Moreover, the population density is so high in our country. Beside this, the carcinoma patients increase day by day due to the illiteracy, poverty, food habited, chain smoking and alcohol. The developing countries like Bangladesh have the limitations of radiotherapy facilities due to the economic constraints.

In this paper, consideration of time, accuracy, number of carcinoma patients and organizational limitations: we have

Md Abdullah Al Mashud (Ph.D. Candidate) is with the Department of Electrical and Electronic Engineering, Islamic University, Kushtia-7003, Bangladesh (e-mail: mashud@ aece.iu.ac.bd).

M. Tariquzzaman is with the Department of Information and Communication Engineering, Islamic University, Kushtia-7003, Bangladesh (corresponding author, e-mail: mtareq@ice.iu.ac.bd).

M Jahangir Alam is with the Department of Radiation Oncology, Ahsania Mission Cancer and General Hospital, Dhaka. Bangladesh.

Tapan Kumar Godder and M. Mahbubur Rahman are with the Department of Information and Communication Engineering, Islamic University, Kushtia-7003, Bangladesh.

proposed a novel method adopting the MC simulation-based treatment planning system for simulation and 2-point measurement technique with adaption of interpolation for measurements. The proposed approach could be more beneficial in cancer treatment in the developing countries. The focus of the present study is to evaluate the accuracy of the TPS for photon calculations using MC simulations in lung irradiation in order to measure accurately the point dose in the lung tumor.

II. Traditional TPR Measurement Method

Basically, TPR is measured using SAD technique, i.e. iso-centric technique which needs huge number of measurement. It is obvious that much measurement is time consuming and may lead to significant human error because of fatigue. Besides, errors may also occur due to complexity of the dosimetry setup. The formula for TPR measurement equation [2] is as in (1)

$$TPR_{(S,Q,d)} = \frac{D_d}{D_{ref}} \quad (1)$$

where Dd is the dose at any depth and Dref is the dose at reference depth

III. Proposed Approach for TPR Measurement

To avoid the complexity of TPR measurement, this paper adopted the following mathematical model [27] for measurement in inhomogeneous medium. This method has been investigated for inhomogeneous medium and found sound results. This research proposed to adapt this approach with MC method-based treatment planning system. The 2-point measurement equation with adaption of interpolation equation is as in (2)

$$TPR(d, FS_A) = \frac{1}{100}.PDD(d, FS_A\frac{X}{Y}, X) \times \left(\frac{Y}{Z}\right)^2 \times \frac{S_{PA}(FS_A, \frac{X}{Y})}{S_{PD}(FS_A, \frac{R}{Y})} \quad (2)$$

where d is the depth in cm for required TPR, FS_A is desired field size for TPR measurement, X= SSD = 100 cm, Y = SSD + d = 100 + d cm, Z = SSD + dr = 100 + reference depth (10 cm), S_{PA} is scattering factor at phantom surface, S_{PD} is scattering factor at depth (d+SSD) and R = SSD + d -dr.

It is easier to perform the PDD measurement using fixed surface to source distance (SSD) technique. Its needs only two values at any two points. Then, we convert these values to TPR. This is also less inclined to error because of less difficulty in the dosimetry setup and measurement. The simulated formula could be used to calculate the TPR values for isocentric treatment, otherwise it would have needed complex and elaborate measurement setup changes under SAD formalism. Therefore, the present work will have a big impact on the quality and safety of radiotherapy in the global arena, particularly in the Third World.

IV. Material and Methods

Tumors in lungs are usually radiated either anteriorly (from the front) or posteriorly (from the back). Here, it is assumed that only a single fraction from any one of the above two directions is used and the phantom was designed accordingly. A rectangular piece of polystyrene having density 1.06 g/cc and size 40cm(l) × 40cm(b) × 1.5cm(h) has been employed for performing the simulation of lung tissue with extensive tumor (large-mass) which is treated as lung tumor tissue equivalent materials [28]. It has been fixed inside the water phantom in such that the bottom of its remains at a depth of 1.5 cm from the water surface. This was done to represent skin, soft tissue above the body fluids. However, the effect would be very small. It also may simulate cancer tumor on the surface of the lung. The water phantom had a dimension of 40cm x 40cm x 40cm. The radiation was applied on the field sizes of 5cm x 5cm and 10cm x10cm. The phantom was made in accordance with the ICRU methodology [29]. The doses were measured along the central beam axis underneath the polystyrene in the phantom. For these dose measurements, an ion chamber with a small volume of 0.125cc was used.

MC based Monaco TPS as a conventional dose calculation system was used in this research. Firstly, a treatment plan has been made for the above inhomogeneous phantom for an SSD of 100 cm which offered isodose distributions for different field sizes at different depths [30]. Then, actual dose measurements were carried out in two dimensions along different horizontal planes corresponding to different depths below the polystyrene inside the water phantom. These were carried out for the specified field sizes and depths as used for the treatment plan. Then, we correlated the treatment plan data and the measured dose to determine the efficacy of our simple inhomogeneous phantom.

The methodology consisted of:

1. The measurements of dose (output) were done with calibrated dosimetry system for different clinical field sizes at 100 cm SSD by using IAEA, TRS-398 protocol [31]. The calibrated ionization chamber was set at reference depth in water phantom. Nowadays, phantoms of other materials are also available for the dosimetry of the teletherapy units but due to equality of density with human tissue, the water phantom has superiority on others [32]
2. IAEA's TRS-398 (2004) protocol [31] was used to obtain the absorbed dose to water at reference depth Z_{ref} in a photon beam of quality Q using the following formula:

$$D_{W,Q}(Z_{ref}) = M_Q \times K_{QQ0} \quad (3)$$

where M_Q= is the reading of the dosimeter with the reference point of the chamber positioned at z_{ref} in accordance with the reference conditions [33].

$$M_Q = M_R \times K_{pol} \times K_s \times N_{DW} \times K_{TP} \quad (4)$$

$$M_R = Elecctrometer\ reading$$
$$K_{pol} = \frac{|M_+|+|M_-|}{2M} \quad (5)$$

K_{Pol} = Change in polarity factor to correct the ionization chamber response on change of polarizing voltage. $|M_+| =$ Electrometer reading at voltage $+V_1$. $|M_-| =$ Electrometer reading at voltage $-V_2$. K_S = Ion recombination correction factor to take two electrometers reading on two voltage settings.

$$K_S = \frac{\left(\frac{V_1}{V_2}\right)^2 - 1}{\left(\frac{V_1}{V_2}\right)^2 - \left(\frac{M_1}{M_2}\right)^2} \quad (6)$$

N_{DW} = Calibration factor of electrometer and ionization chamber for absorbed dose to water. K_{TP} = Temperature and pressure correction factor and

$$K_{TP} = \frac{273.2+T}{273.2+T_0} \times \frac{P_0}{P} \quad (7)$$

where P_0 and T_0 are the reference values of pressure and temperature respectively and were taken as 101.3 kPa and 20 °C. $K_{Q,Qo}$ is a chamber-specific factor which corrects for the difference between the reference beam quality Q_o and the actual quality being used, Q.

The absorbed dose rate to water at the depth of dose maximum, Z_{max} is

$$D_{W,Q}(Z_{max}) = \frac{D_{w,Q}(Z_{ref})}{TPR(Z_{ref})} \times 100 \quad (8)$$

3. The percentage variation for event of dosimetry had been calculated by comparing the output obtained by the measured value and the simulated value.

V. Results and Discussion

The 6-MV photon beam direction was fixed along the zero angles of the gantry and collimator. The ion chamber was fixed perpendicular to the central axis radiation beam. The field sizes of the beam were fixed at 5 cm×5 cm, 10 cm×10 cm and 12 cm×12 cm and the central axis depth doses recorded in units of cGy/MU for different depths. The depth doses were measured at 0.2 cm increment up to 20 cm in water with and without polystyrene. The simulated TPS of depth dose data at 0.2 cm increment up to 20 cm has been collected.

A. Perturbation Calculation

The perturbation measurement and calculation between the polystyrene values and the corresponding without polystyrene values were determined for various field sizes and the results are tabulated in Table I.

TABLE I
Perturbation Calculation between Polystyrene and without Polystyrene

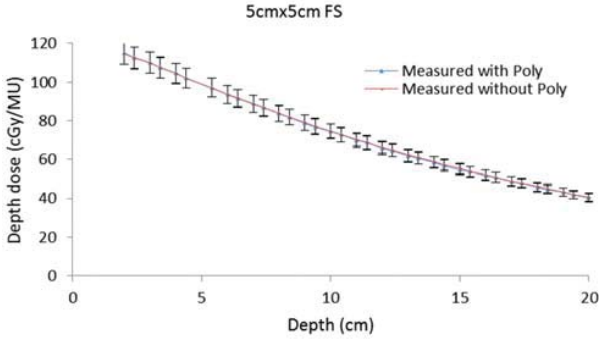
Field size	Measured	TPS Cal.
5cm×5cm	-0.86 to0.15	0.05 to -0.91
10cm×10cm	-2.05 to 2.02	-0.12 to -1.68
12cm × 12cm	-3.01 to 0.60	-0.22 to -2.08

B. Depth Dose Measurement

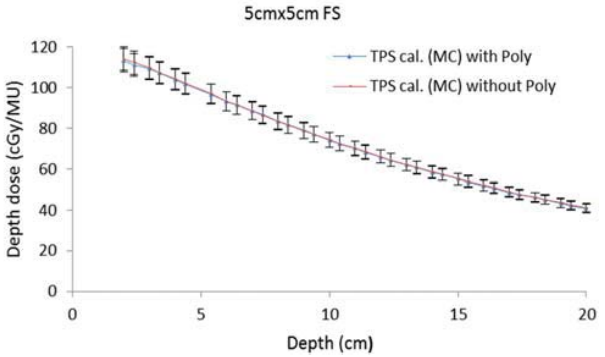
The measured depth dose values are recorded and compared with the TPS simulated data for different field sizes. The graphical representation of depth dose of with polystyrene and without polystyrene of measured, and TPS simulated values are shown in Fig. 1. The treatment planning system has calculated radiation dose theoretically and using some mathematical model for inhomogeneous fields by considering the radiation basic beam data of homogeneous fields. The other correction factors were also calculated theoretically like phantom scattered correction factors, collimator scattered correction factors, tissue density correction factors and other perturbation factors [22]. However, for the in-phantom measurements, all correction factors are included with the dosages.

TABLE II
Percentage of Deviation for Measured and TPS Simulated Values with and without Polystyrene

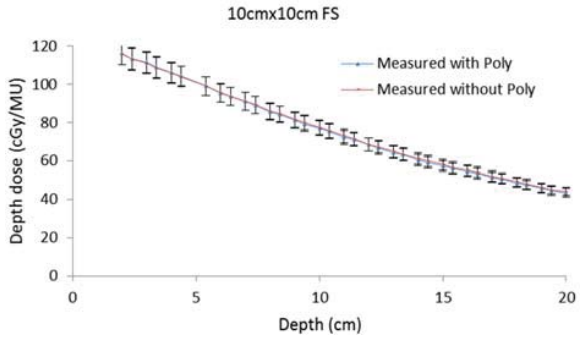
Field size (cm^2)	Measured (betⁿ polystyrene and without polystyrene)			TPS simulated (betⁿ polystyrene and without polystyrene)		
	Min difference	Max difference	Mean difference	Min difference	Max difference	Mean difference
5×5	-1.20%	-0.60%	-0.27%±0.44%	-0.98%	0.44%	-0.29%±0.39%
10×10	-1.40%	0.36%	-0.62%±0.54%	-1.55%	0.48%	-0.57%±0.40%
12×12	-1.91%	0.94%	-0.84%±0.76%	-0.19%	-0.13%	-0.94%±0.41%



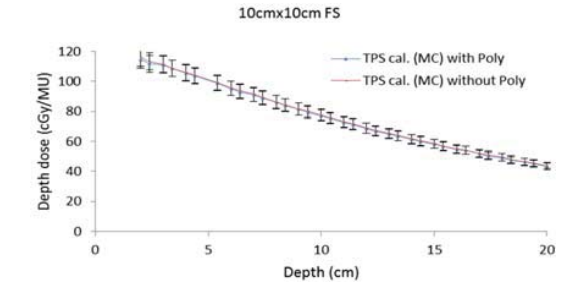
(a)



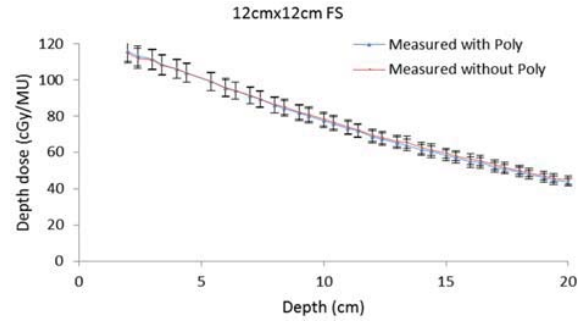
(b)



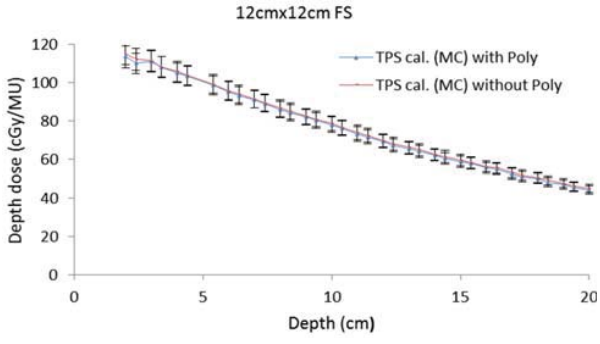
(c)



(d)



(e)



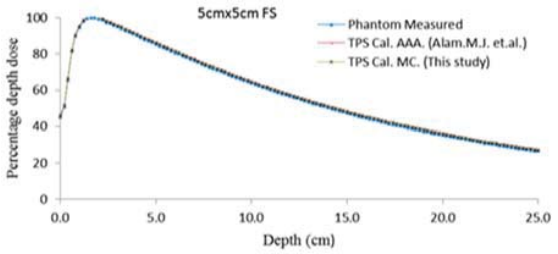
(f)

Fig. 1 Comparison of depth dose (Measured and TPS simulated) with polystyrene (inhomogeneous) & without polystyrene (homogeneous) (a) measured value of field size 5 cm^2 (b) simulated value of field size 5 cm^2 (c) measured value of field size 10cm^2 (d) simulated value of field size 10 cm^2 (e) measured value of field size 12 cm^2 (f) simulated value of field size 120 cm^2

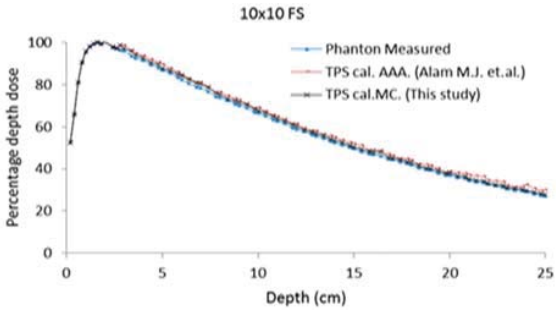
It was observed from Fig. 1, that the deviations of measured and TPS simulated values were very small. The percentage of deviations of measured (between polystyrene and without polystyrene) and TPS simulated (between polystyrene and without polystyrene) have been carried out and summarized in Table II. These results are very satisfactory since variation up to 3.15% is considered acceptable [34]. Therefore, this shows a good agreement of dose calculation and QA of control data with and without polystyrene in both cases of measurement and treatment planning system.

C. PDD Measurement

The dose rate measurements for the different field sizes are shown in Fig. 2. The corresponding TPS calculated data are also shown in the same graphs. It also compared with another data that was simulated by AAA algorithm [25] in the same graph respectively. The percentage of deviation of PDD between TPS simulation and phantom measure is tabulated in Table III. The difference between the measured and calculated TPS values for the above field sizes can be observed from these graphs, which are very small. The differences between the measured values and the corresponding TPS calculated values were determined for various field sizes, and these were in the range of -0.020% to 2.154%. This result is very satisfactory since variation up to 3.5% is considered acceptable [31]. The average %variation obtained 1.337±0.522 and 1.484±0.725 for 5 cm×5 cm and 10cm×10cm field size, respectively. Therefore, this shows sound results of dose calculation and QA of treatment planning. Application of such techniques will be beneficial to patients for radiotherapy treatment and dose delivery to the targeted area.



(a)



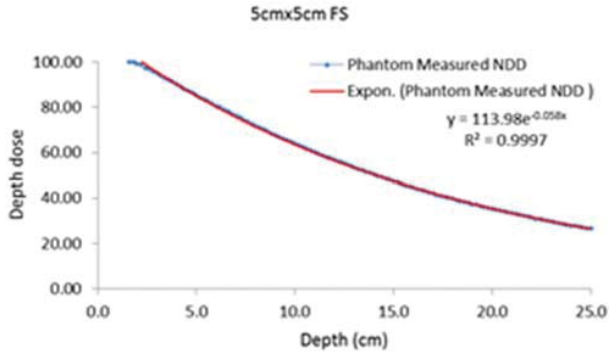
(b)

Fig. 2 Percentage depth dose for a 6MV photon beam Monaco TPS and measured using a 0.125 cc Farmer ionization chamber in an inhomogeneous phantom at SSD=100 (a) field size 5 cm×5 cm and (b) 10 cm×10 cm

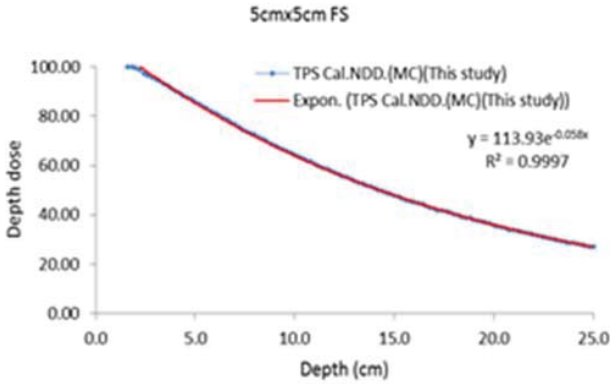
TABLE III
PERCENTAGE DEVIATION OF PDD BETWEEN TPS SIMULATED AND PHANTOM MEASURED VALUES

Field size	Proposed Approach		Alam et al. [25]	
	Min difference	Max difference	Min difference	Max difference
5cm×5cm	0.270%	2.154%	0.892%	3.592%
10cm×10cm	-0.020%	2.142%	0.008%	3.38%

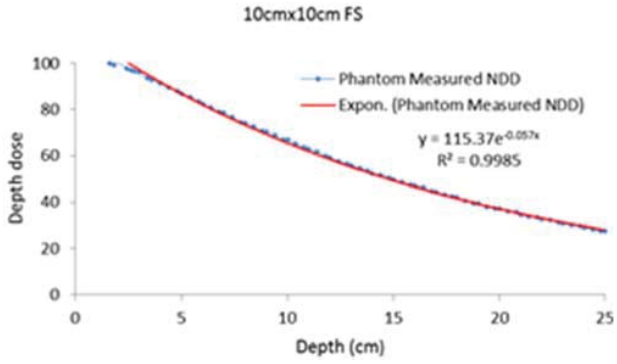
The curve fitting of measured and simulated TPS values of field sizes 5 cm×5 cm and 10 cm×10 cm is shown in Fig. 3. From the curve, it shows the differences between measured values and TPS simulated values vary within ±2%. The results show good agreement with the recommendations. The R^2 values are also summarized in Table IV.



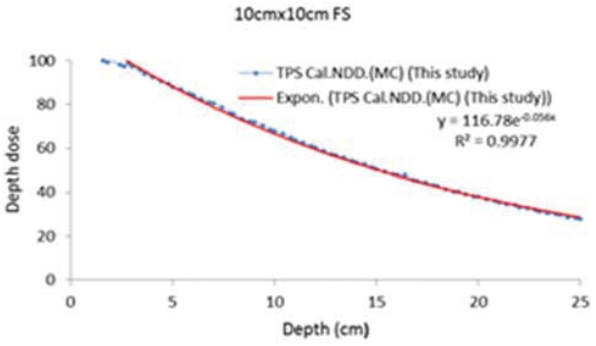
(a)



(b)



(c)



(d)

Fig. 3 Curve fitting of measured and TPS simulated percentage depth dose for a 6-MV photon beam (a) measure value for field sizes 5 cm×5 cm (b) simulated value for field size 5 cm×5 cm (c) measure value for field sizes 10 cm×10 cm and (d) simulated value for field size 10 cm×10 cm

TABLE IV
CURVE FITTING OF R^2 VALUES FOR DIFFERENT FIELD SIZES OF 6 MV PHOTON BEAM

Field sizes	Measured	Simulated by MC (This study)	Simulated by AAA [25]
5cm×5cm	0.9997	0.9997	0.9998
10cm×10cm	0.9985	0.9977	0.9974

VI. CONCLUSION

For optimum treatment of cancer, the radiation dose must be planned and delivered with a high degree of accuracy. The international commission on radiation units and measurements (ICRU) recommends that the dose should be delivered with a deviation not greater than 5% with respect to a treatment plan, but accuracy of ±3% to ±3.5% in the overall process has been recommended [35], [36]. In this study, the overall deviation between the measured and simulated data was within or near to 2%. The maximum deviation for a very few data points was 3.29% what is beyond the recommended limit, but still within the ICRU recommendation. Therefore, the proposed approach is more efficient and will be beneficiary the developing countries like Bangladesh.

REFERENCES

[1] Ganesh N, Wilbert C, Niko P, Sotirios S. Comparison between measured tissue phantom ratio values and calculated from percent depth dose with and without peak scatter correction factor in a 6 MV beam. Int J Cancer Ther Oncol 2015:3(2):03024.
[2] Gibbons et al.: Monitoring unit calculations for photon and electron beams. Med. Phys. 41(3), March 2014: 031501-1.
[3] Brahme A, Lax I, Andreo P. Electron beam dose planning using discrete Gaussian beams. Mathematical background. Acta Oncol. 1981;20(2):147–158.
[4] Hogstrom KR, Mills MD, Almond PR. Electron beam dose calculations. Phys Med Biol. 1981;26(3):445–459.
[5] Mohan R, Chui C, Lidofsky L. Differential pencil beam dose computation model for photons. Med Phys.1986;13(1):64–73.
[6] Ulmer W, Pyyry J, Kaissl W. A 3D photon superposition/convolution algorithm and its foundation on results of Monte Carlo calculations. Phys Med Biol. 2005;50(8):1767–1790.
[7] Boyer A, Mok E. A photon dose distribution model employing convolution calculations. Med Phys. 1985;12(2):169–177.
[8] Ahnesjö A. Collapsed cone convolution of radiant energy for photon dose

calculation in heterogeneous media. Med Phys. 1989;16(4):577–592.
[9] Antonella Fogliata et. al,. Dosimetric validation of the anisotropic analytical algorithm for photon dose calculation: fundamental characterization in water. Phys. Med. Bio. 2006; 51:1421-1438.
[10] Fippel M, Kawrakow I, Friedrich K. Electron beam dose calculations with the VMC algorithm and the verification data of the NCI working group. Phys Med Biol. 1997;42(3):501–520.
[11] Kawrakow I, Fippel M. Investigation of variance reduction techniques for Monte Carlo photon dose calculation using XVMC. Phys Med Biol. 2000;45(8):2163–2183.
[12] Hartmann Siantar CL, Walling RS, Daly TP, et al. Description and dosimetric verification of the PEREGRINE Monte Carlo dose calculation system for photon beams incident on a water phantom. Med Phys. 2001;28(7):1322–1337.
[13] Neuenschwander H, Mackie TR, Reckwerdt PJ. MMC—a high-performance Monte Carlo code for electron beam treatment planning. Phys Med Biol. 1995;40(4):543–574.
[14] Fippel M. Fast Monte Carlo dose calculation for photon beams based on the VMC electronalgorithm. Med Phys.1999;26(8):1466–1475.
[15] Kawrakow I. Proceeding of the Monte Carlo 2000 Meeting Lisbon. In: Kling A, Barao FJC, Nakaqawa M, et al., editors. Advanced Monte Carlo for radiation physics, particle transport simulation and applications. Berlin (Germany): Springer; 2001: 229–236
[16] Paelinck L, Reynaert N, Thierens H, De NW, De WC. Experimental verification of lung dose with radiochromic film: comparison with Monte Carlo simulations and commercially available treatment planning systems. Phys Med Biol. 2005;50(9):2055–2069.
[17] Werner BL, Das IJ, Salk WN. Dose perturbations at interfaces in photon beams: secondary electron transport. Med Phys. 1990;17(2):212–226.
[18] Van DJ, Barnett RB, Cygler JE, Shragge PC. Commissioning and quality assurance of treatment planning computers. Int J Radiat Oncol Biol Phys. 1993;26(2):261–273.
[19] Brahme A, Chavaudra J, Landberg T, et al. Accuracy requirements and quality assurance of external beam therapy with photons and electrons. Acta Oncol. 1988;27(1 Suppl):1–76.
[20] Jeff Craig, Mike Oliver, Adam Gladwish, Matt Mulligan, Jeff Chen and Eugene Wong. Commissioning a fast Monte Carlo dose calculation algorithm for lung cancer treatment Planning. Journal of Applied Clinical Medical Physics. 2008; 9(2): 83-93.
[21] Fox R. A. & Webb S. A comparison of inhomogeneity correction methods with Monte Carlo data. Australasian Physical Sciences in Medicine, 2 – 8, no.85, 452-462. Nov/Dec. 1979.
[22] Chow JC, Wong E, Chen JZ, Van Dyk J., Comparison of dose calculation algorithms with Monte Carlo methods for photon arcs, Med. Phys. 30(10), Am. Assoc. Phys. Med, 2003, 30(10), 2686-2694.
[23] George X Ding, Dennis M Duggan, Charles W Coffey, Parvaneh Shokrani and Joanna E Cygler. First macro Monte Carlo based commercial Dose calculation module for electron beam Treatment Planning, new issues for clinical consideration. Phys. Med. Biol. 51, 2781-2799, 2006.
[24] Hideharu Miura et al.: Evaluation and Commissioning of Commercial Monte Carlo Dose Algorithm for Air Cavity. Int J Medical Physics, Clinical Engineering and Radiation Oncology, 2014: 3: 9-13.
[25] Alam M.J. et.al. Calculation Accuracy and Quality Assurance of Computer Aided Radiation Therapy Planning. Ph.D., Thesis, University of Dhaka, 2012.
[26] E. G. A. Aird et.al, Central Axis Depth Dose data for Use in Radiotherapy. British Journal of Radiology Supplement-25.
[27] M. J. Alam, K. S. Rabbani, S. M. A. Husain, A. Kabir and T. Bagi. A modified formula for defining tissue phantom ratio of photon beams. Bangladesh Med Res Counc Bull 2007; 33: 92-921996.
[28] M. Jahangir Alam, M. Ashrafur Rahman, Khandoker Siddique-E. Rabbani. Dosimetric Accuracy Using the New Mathematical Tools for Inhomogeneous Denser Medium. International Journal of Clinical and Experimental Medical Sciences. Vol. 3, No. 2, 2017, pp. 23-29. doi: 10.11648/j.ijcems.20170302.12.
[29] J R Greening. Fundamentals of Radiation Dosimetry, Second Edition: Institute of Physics Pub., in collaboration with the Hospital Physicists' Association. 1992; p-176.
[30] Webb S. & Fox R. A,. Verification by Monte Carlo methods of power law tissue-air ratio algorithm for inhomogeneity corrections in photon beam Dose calculations. Phys. Med. Biol. 1980, 25(2); 225-240.
[31] Pedro Andreo et.al,. Absorbed Dose Determination in External Beam Radiotherapy. An International Code of Practice for Dosimetry Based on Standards of Absorbed Dose to Water. Technical Report Series no. IAEA TRS-398, 2004.
[32] S. A. Memon, N. A. Laghari, F. H. Mangi, F. Ahmad, M. M. Hussain, S. Palijo, N. Jhatyal and A. Adeel. “Analysis and verification of percent depth dose and tissue maximum ratio for Co-60 gamma ray beam.” Worl App Sci J 2015;33(1):109-13. http://dx.doi.org/10.5829/idosi.wasj.2015.33.01.9261.
[33] Samuelsson A. and Johansson K.A., Dose Accuracy Check of the 3-D Electron Beam Algorithm in Cad Plan. Sahlgrenska University Hospital, September 1995.
[34] William L. Saylor, M. S., D. A. B. R. and Thomas E., Dosage Calculations in Radiation Therapy, Ames, B. S., R. T. Urban & Schwarzenberg, 1979.
[35] Webb S., The Physics of Three-Dimensional Radiation Therapy: Conformal Radiotherapy, Radiosurgery, and Treatment Planning, (Ph. D.) p- 373 Jan 1, 1993.
[36] M. R. Sontag and J. R. Cunningham. The equivalent tissue–air ratio method for making absorbed dose calculations in a heterogeneous medium. Radiology 129, 787–794(1978).