A Novel Dose Distribution Approach using Monte Carlo Simulation in Dosimetric Accuracy Calculation for Treating the Lung Tumour

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

***Keywords*—** Lung tumour, 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 delivering the maximum dose to the cancerous tissues and minimum dose to the normal tissues and risk organs, and save the normal tissue. This approach is involved in variety of cancer treating research.

One of the most significant parameter in dosimetric analysis is Tissue Phantom Ratio (TPR) measurement [1]. Instead of using the available conventional expression we can choose 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 very convenient to measure the percent depth dose (PDD) but it is inconvenient for patients, since, basically 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 huge number of measurement what time consume.

Besides, several commercial treatment planning systems uses 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 Monte Carlo Methods [10]-[15]. For high calculation speed, the CCC method is widely used in commercial treatment planning systems. However, the differences greater than 5% have found at the materials for different densities such as between lung and tissue [16]-[17]. The 5% difference is deemed unacceptable for better dosimetry to deliver the dose to the patient [18]-[19]. The dose calculation speed is comparatively slow in Monte Carlo method however; it has better accuracy comparing to the others algorithms [20]. Monte Carlo based calculation focus the inhomogeneity correction [21]. Photon beam dose calculation using Monte Carlo methods [22] has adapted the American Association of Physicist of Medicine. The Monte Carlo base dose calculation has suggested [23] for commercial treatment planning systems and for air cavity measurement [24]. Dosimetric accuracy calculation and quality assurance has carried out using AAA algorithm [25], however, the difference is found less than 5% but very greater than 2%. The accuracy is recommended less than 2% for better treatment [26].

However, the population density is very high in our country. Besides, the carcinoma patients increases 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 constrains.

In this paper, consideration of time, accuracy, number of carcinoma patients and organizational limitations: we have proposed a novel method adopting the Monte Carlo 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 Monte Carlo simulations in

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lung irradiation in order to measure accurately the point dose in the lung tumour.

II. TRADITIONAL TPR MEASUREMENT METHOD

Basically, Tissue Phantom Ration (TPR) is measured using Source to Axis Distance technique (SAD setup) 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, we have adopted the following mathematical model [27] for measurement in inhomogeneous medium. This method have investigated for inhomogeneous medium and found sound results. We proposed to adapt this approach with Monte Carlo 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, FSA 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), SPA is scattering factor at phantom surface, SPD is scattering factor at depth (d+SSD) and R = SSD + d - dr.

It is easier to perform the percent depth dose (PDD) measurement using fixed surface to source distance (SSD) technique. We need only two values at any two points. Then convert this value 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 would have needed complex and elaborate measurement set up changes under Source to Axis Distance (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

Tumours 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. To simulate lung tissue having an extensive tumour (large mass) we used a rectangular piece of polystyrene of density 1.06g/cc and of size 40cm(l) × 40cm(b) × 1.5cm(h). We fixed it inside the water phantom such that the bottom of the polystyrene block remains at a depth of 1.5 cm from the water surface. This was done to represent skin, soft tissue above the body fluids. The effect would however, be very small. It also may simulate cancer tumour 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 [28].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.

Monte Carlo based Monaco TPS as a conventional dose calculation system was used in the present work. Firstly, we made a treatment plan for the above inhomogeneous phantom for an SSD of 100cm which gave isodose distributions for different field sizes at different depths [29]. 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 100cm SSD by using IAEA, TRS-398 protocol [30]. 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 [31]

2. IAEA's TRS-398 (2004) protocol [30] 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 [32].

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

$$M_R = Elecctrometer\ reading$$

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

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 electrometer 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 6MV 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 5cm×5cm, 10cm×10cm and 12cm×12cm 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. It has been collected the TPS simulated of depth dose data at 0.2 cm increment up to 20 cm.

A. Perturbation Calculation

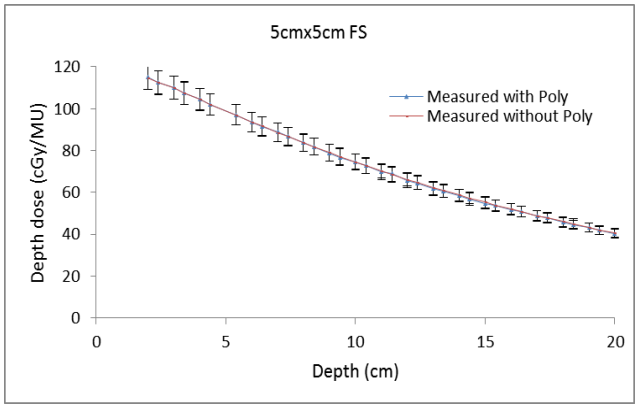
The perturbation measurement and calculation between the polystyrene values and the corresponding without polystyrene values were determined for various field sizes and the results are tabulated in Table I.

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 compare 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 dosage 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



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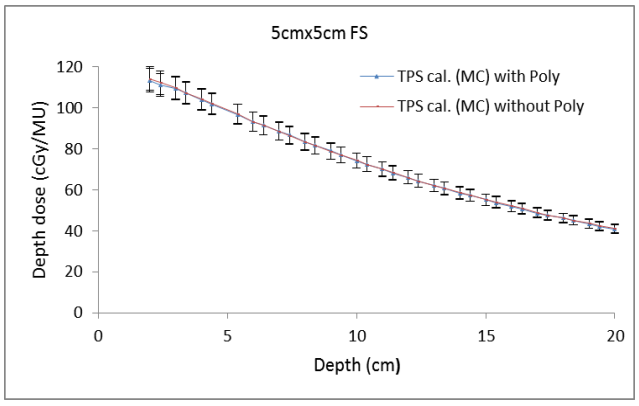

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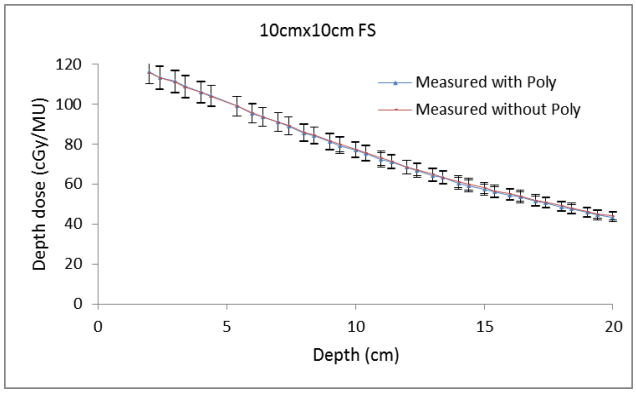

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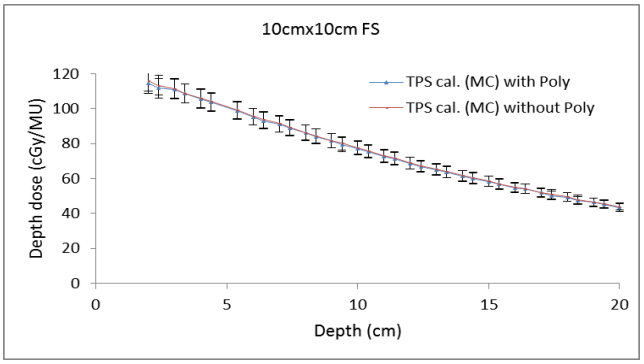

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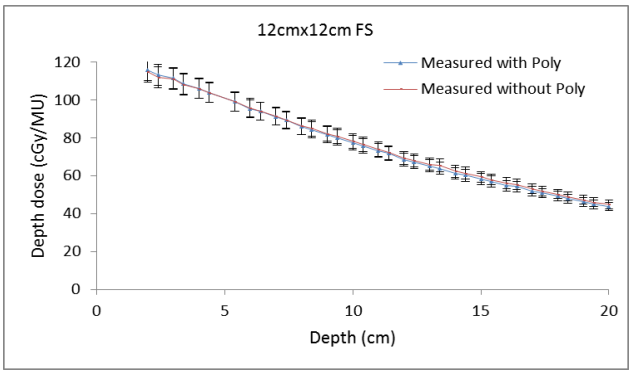

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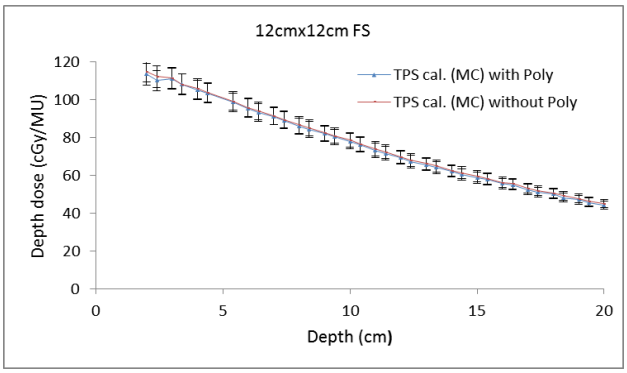


(f)

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

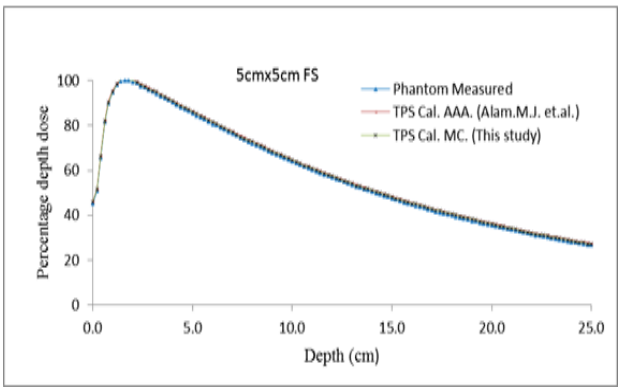
TABLE II
PERCENTAGE OF DEVIATION FOR MEASURED AND TPS SIMULATED VALUES WITH AND WITHOUT POLYSTYRENE

Field size (cm^2)	Measured (bet^n polystyrene and without polystyrene)			TPS simulated (bet^n 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%

These results are very satisfactory since variation up to 3.15% is considered acceptable [33]. 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 is compare with another data what is 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 [30]. The average %variation obtained 1.337±0.522 and 1.484±0.725 for 5cm×5cm 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.



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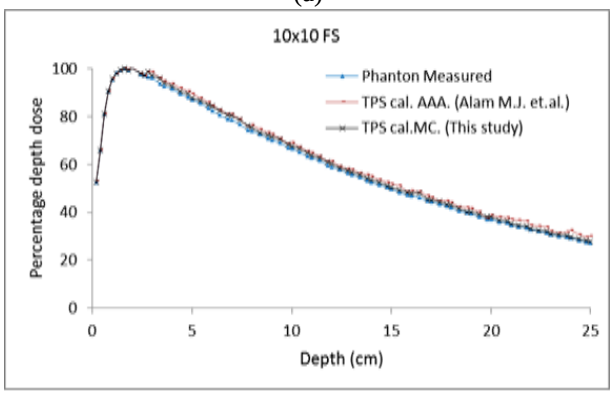


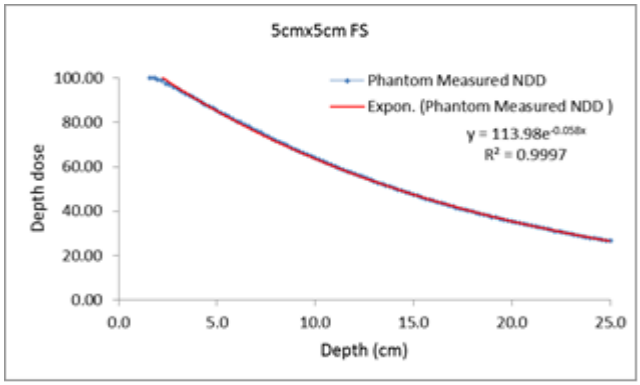
(b)

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 5cm×5cm and (b) 10cm×10cm

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

Field size	Proposed Approach		Alam M.J 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 TPS simulated values of field sizes 5cm×5cm and 10cm×10cm are 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.



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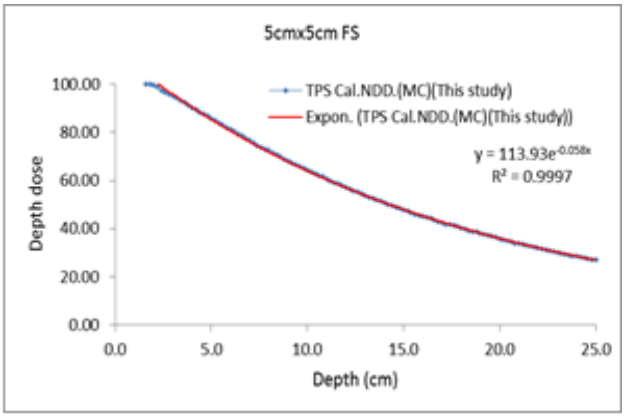

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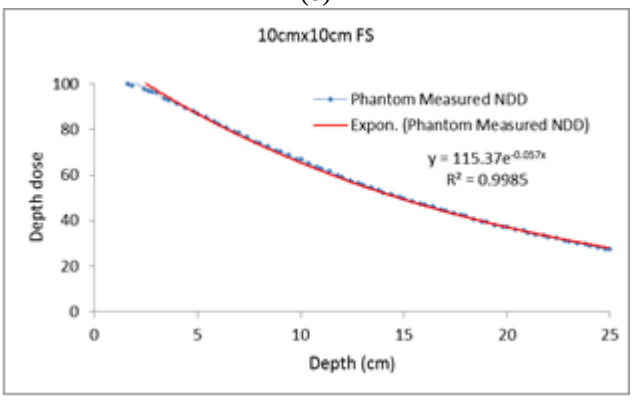

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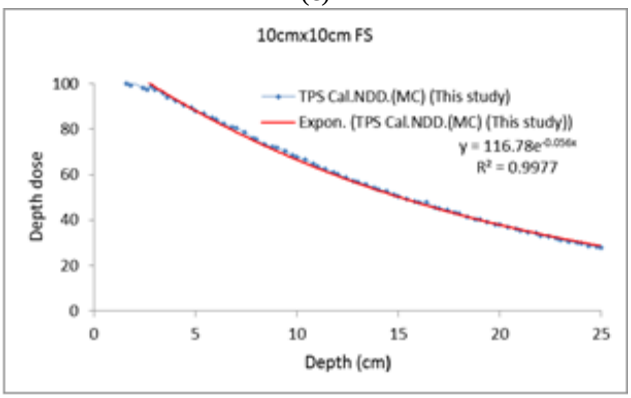

(d)

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

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

Field sizes	Measured	Simulated by Monte Carlo (This study)	Simulated by AAA [22]
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 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 [34]-[35]. In our study the overall deviation between 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.

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