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Original article

Verification the Accuracy of Anisotropic Analytical Algorithm (AAA)

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Purpose: The aim of this study is to verify the dose calculation accuracy of anisotropic analytical algorithm (AAA) and evaluation of this accuracy in comparison to Pencil Beam Convolution (PBC) especially in low density tissue heterogeneity (lung tissue) through comparing calculated doses with the measured doses through thermoluminescent dosimeters (TLD) for 6 MV photon beam, in order to evaluate the accuracy of the AAA and comparing it with the accuracy of PBC.
Materials and Methods: Selected field sizes were measured in water phantom then compared with calculated ones through two different algorithms PBC and AAA. Rando humanoid phantom was imaged using CT then it was transferred to the Eclipse planning system where similar plan of single direct field calculated two times one through AAA and the other through PBC then dose measured within phantom using TLD in the lung region then measured and calculated data are compared.
Results: There was good agreement between measured fields in water phantom and calculated AAA and PBC measurements but AAA was more accurate than PBC. Also AAA was better than that of PBC however maximum dose deviation of AAA was -4.5% compared to -6.7% for PBC at the same point.
Conclusion: AAA measurements were more accurate compared to those calculated with PBC in low density heterogeneity.

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INTRODUCTION

Accuracy in treatment planning systems (TPSs) has always been a concern in modern radiotherapy. An accurate determination of the absorbed dose in radiation therapy is very important. For Planning Target Volume (PTV) plan, according to the International Commission of Radiation Units and Measurements (ICRU) 50 (ICRU 1993) the deviation in the dose delivered should be kept within -5% and +7% of the prescribed dose in the plan[1]. To achieve this goal all involved steps should be minimized regarding uncertainties. The work of dose calculations can be done by using Treatment Planning Systems (TPS). The ICRU 42 (ICRU 1987) states that computer-produced dose distributions can be considered accurate enough if they differ from relative dose measurements by less than 2%, or 0.2cm in position of isodose curves in very steep dose gradients[2].

Inadequate ability of TPS to correctly estimate dose distribution for the tissue heterogeneity is one of the error sources in radiotherapy process which leads to error in dose delivery to targeted PTV. Lung tissue is one of the most important low-density heterogeneities that impact upon the success of a radiation therapy. That heterogeneity deserves special consideration also the breathing motion of the lungs is a factor increasing the complexity during the therapeutic dose administration.

Breitman *et al*[3] compared the performance of AAA for Varian linear accelerator with measurements performed at two institutions using 6 MV and 15 MV beams. The TG-53 (AAPM report 1998) evaluation regions and criteria were used to evaluate profiles measured in a water phantom for a wide variety of clinically relevant beam geometries. The Total Scatter Factor (TSF) for each of these geometries was also measured and compared against the results from the AAA. At one institute, TLD measurements were performed at several points in the neck and thoracic regions of a Rando phantom; at the other institution, ion chamber measurements were performed in a CIRS inhomogeneous phantom. The phantoms were both imaged using CT and the dose was calculated using the AAA at corresponding detector locations. Evaluation of measured relative dose profiles revealed that 97%, 99%, 97% and 100% of points at one institute and 96%, 88%, 89% and 100% of points at the other institution passed TG-53 evaluation criteria in the

outer beam, penumbra, inner beam, and buildup regions respectively [3].

Gagné and Zavgorodni[4] compared the performance of AAA to that of Pencil Beam Convolution (PBC) in an extreme (C shaped, horizontal and vertical boundaries) water–lung interface phantom. Monte Carlo (MC) calculated dose distributions were used as benchmarks in the comparison. AAA was considerably more accurate than PBC with the standard deviation of the dose differences within a region encompassing the lung block reduced by a factor of 2 and more. AAA calculations for the small 4×4 cm² 18 MV beam, which is poorly modeled by PBC (dose differences as high as 16.1%), provided the same accuracy as the PBC model of the 6 MV beams commonly acceptable in clinical situations[4].

Bragg *et al*.[5] determined the implications of the use of the AAA for the production and dosimetric verification of Intensity Modulated Radiation therapy (IMRT) plans for treatments of the prostate, parotid, nasopharynx and lung, 72 IMRT treatment plans produced using the PBC algorithm was recalculated using the AAA and the dose distributions compared. They concluded that, all AAA calculations were within 3% or 3.5mm distance to agreement of the measured doses. Since the reported differences in the case of AAA were, in general, less than the disagreements in case of PBC algorithm, they recommended that the AAA should be used in preference to the PBC algorithm for treatments involving low density tissue but this may necessitate re-evaluation of plan acceptability criteria. There was excellent agreement between the AAA and verification measurements for all sites[5].

The inhomogeneity correction capabilities of AAA was examined by Robinson.[6] The investigator used planar geometries consisting of three layers of solid water the upper layer was 4 cm thickness, the lowest layer was 20 cm thickness, while the middle layer thickness was variable as he used this layer to investigate the heterogeneity capability of AAA by replacing the homogenous layer of solid water by, one at time, materials of equivalent thickness, but of different composition. It was found that the AAA over predicted dose beyond low-density regions and under predicted dose distal to volumes of high density[6].

da Rosa *et al*.[7] investigated the influence of lung heterogeneity inside a soft tissue phantom on PDD. PDD curves were obtained experimentally using LiF:Mg,Ti (TLD-100) and by applying Eclipse TPS algorithms: Batho, modified Batho (M-Batho or BMod), equivalent TAR (E-TAR or EQTAR), and AAA for a 15 MV photon beam and field sizes of 1 × 1, 2× 2, 5 ×5 and 10 × 10 cm². Monte Carlo simulations were performed using the DOSRZnrc user code of EGSnrc. The experimental results agreed with Monte Carlo simulations for all irradiation field sizes. Comparisons with Monte Carlo calculations showed that the AAA algorithm provided the best simulations of PDD curves for all of the investigated field sizes. However, even that algorithm couldn't accurately predict PDD values in the lung for field sizes of 1 × 1 and 2 × 2 cm². An overdose in the lung of about 40% and 20% is calculated by the AAA algorithm close to the soft tissue/lung interface for 1 × 1 and 2 × 2 cm² field sizes, respectively[7].

A study investigated the ability of AAA and PBC to calculate the dose in deep-seated water equivalent tissue beyond high density heterogeneity interface by comparing the computed data with measured ones. Combination of solid water, Poly Vinyl Chloride (PVC) and Styrofoam were manufactured to simulate a phantom. Initially depth dose measurements at 1 cm interval along central axis in homogenous medium which represent a benchmark were done. Data were acquired for variable field sizes (5 x 5, 10 x 10 and 20 x 20 cm2) through cylindrical ionization chamber at selected depths beyond high density heterogeneity interface. They found that AAA showed a deviation of 5.8% while BPC showed a deviation of 6.7%. They concluded that AAA is more accurate than PBC for dose calculation in treating deep seated tumors beyond high density heterogeneity[8].

The aim of this study is to evaluate the dose calculation ability of anisotropic analytical algorithm (AAA) in low density heterogeneity (lung tissue) through comparing calculated doses with the measured doses through Thermoluminescent Dosimeters (TLD) for 6 MV photon beam, in order to evaluate the performance of the AAA and comparing it with performance of PBC.

MATERIALS AND METHODS

Materials

Linear Accelerator

The ELEKTA (Precise) Linac was used in this study. It can treat fields ranging in size from 4 x 4 cm^2 up to 40 x 40 cm^2 at a 100 cm Source to Skin Distance (SSD). The dose rate for stationary therapy is variable from minimum dose rate of 50 MU / min to 1000 MU /min. The linear accelerator was calibrated so that a square symmetric field 10 x 10 cm² using 100 MU will deliver an absorbed dose of 1 Gy at 10 cm depth in water with SSD of 90 cm.

Treatment Planning System (TPS)

Eclipse planning system (version 8.6.17) is loaded with AAA and PBC with the same version of the planning system was used in this study. All calculations were performed using AAA at 2.5mm calculation grid. Then they are recalculated using PBC at the same grid size for comparison.

Water Phantom

A computerized welhöfer blue water phantom with software WP700 version 3.5 was used for acquiring beam data. The system consists of a cubic water tank of side length of 48 cm and two cylindrical water proof chambers each of sensitive volume of 0.147 cm and wall thickness of 0.04 cm.

Humaonid Rando

Alderson RANDO phantom was used in this study. The phantom is 175 cm tall and 73.5 kg weight. It consists of 34 slices each of 2.5 cm thickness. Each slice is perforated to accommodate the TLD chips. The phantom is made of tissue-equivalent material. The skeleton is highly detailed polymer moldings to simulate the shape, mass, density and attenuation coefficients of cortical bone and spongiosa. The lung is molded from synthetic foam.

Computed Tomography (CT)

The multislice CT used in this work is General Electric (UK); light speed RT. This CT can record four CT images per rotation. The CT unit is provided with a flat carbon fiber table to simulate the linac couch.

Thermo-Luminescent Dosimeter (TLD) system

LiF (700) chips were used in this study for measuring dose in the humanoid phantom. The chip material is LiF enriched with Mg and Ti impurities [Each chip is 0.45 cm (diameter) x 0.08 cm (thickness)]. The effective atomic number is 7.4. The system also includes a Rados TLD reader unit which is used to read the dose measured by TLD chips.

Methods

Verification of AAA: depth dose and beam profile measurements

Phantom of 40 x 40 x 40 cm^3 was created as 3D structure on Eclipse planning system, which was defined as body structure (CT value=0) and used for computation of depth doses and beam profiles for calculated field sizes for both AAA and PBC algorithms then comparing these Percentage Depth Doses (PDDs) and beam profiles with measured data acquired through water phantom at SSD=100cm for 6 MV photon beam from linear accelerator. PDDs were measured along central axis of each field all depth doses are normalized to 100% at depth of maximum dose (1.5 cm) for the 6 MV photon beam. Beam profiles were measured in the central transverse plane (X-Z plane) using the welhöfer water phantom. The dose profiles were measured and calculated at depths of 1.5, 5 and 15cm for the 6 MV photon beam.

TLD (LiF 700) preparation procedures

A new batch of LiF (700) chips was used. The chips were annealed before use by heating up to 400ºC for one hour and for two hours at 100ºC then left in the oven to cool to room temperature. All chips were then irradiated to 1 mGy using RADOS irradiator. Then sensitivity factor for each TLD (LiF 700) chip was measured by dividing the TLD (LiF 700) reader count by the given dose. The average factor is calculated for all chips and considered as 100%. All chips with sensitivity factors exceeding ±2% were omitted.

Evaluation of AAA

Humanoid rando phantom was CT scanned with slice spacing of 2.5 mm, these series transferred to TPS through DICOM network. Risk structures were delineated for chest region including right lung, left lung, heart and spinal cord. A plan of single isocentric direct field of 10 x 15 cm2, a dose of 100 Monitor Units; doses at certain points (representing the positions of TLD chips) are calculated and recorded. Plan isocenter position was checked through simulation step in which anterior and lateral DRR views were compared to those taken by simulator in order to make sure that the position of isocenter is accurate (Figure. 1). TLD chips inserted within their predefined positions in which two crystals were inserted in each position then phantom was irradiated to the previously mentioned field (Figure. 2). Then crystals were collected then dose is measured through TLD reader. The previous step is repeated 3 times for accuracy.

Dose assessment

1. The dose of each chip was then calculated using the previously calculated sensitivity factor.

2. The measured and calculated doses were compared. For TLD measurements the percentage difference between measured and calculated doses is calculated as follows:

% Dose difference = 100 x (Dose measured - Dose calculated) / Dose measured)

RESULTS

Comparison between standard measurements

(Figure. 3): A, B, C, D, E, F, G and H shows PDDs for field sizes 5 x 5 cm^2, 10 x 10 cm^2, 15 x 15 cm^2, 25 x 25 cm^2, 5 x 10 cm^2, 10 x 15 cm^2, 10 x 20 cm^2 and 10 x 25 cm^2 respectively comparing between measured PDD and calculated PDDs through AAA and PBC algorithms and showed good agreement between measured field sizes and calculated ones using PBC and AAA.

(Figure. 4): A, B, C, D, E, F, G and H shows measured and calculated beam profiles for field sizes 5 x 5 cm^2, 10 x 10 cm^2, 15 x 15 cm^2, 25 x 25 cm^2, 5 x 10 cm^2, 10 x 15 cm^2, 10 x 20 cm^2 and 10 x 25 cm^2 respectively. There was good agreement between measured and calculated beam profiles. However

calculated profiles of PBC underestimated the dose at profile shoulder with maximum deviation of 0.9%, 0.8%, 1.5%, 1.5%, 0.2%, 0.7%, 1.5% and 1.4% respectively at 1.5 cm depth. While AAA overestimated the dose at profile shoulder with deviation of 1%, 0.9%, 1%, 1%, 0.2%, 0.3%, 0.2% and 0.2% respectively. Those deviations may be due to inability of AAA to account properly for the attenuations in the beam way.

Measurements of tissue heterogeneity

Table 1 shows results of the measured doses (with TLD), and the calculated (by Eclipse TPS) relative doses. It is obvious that AAA is more accurate than PBC especially at interface regions such as point 1 representing bone air interface and point 2 representing air soft tissue interface.

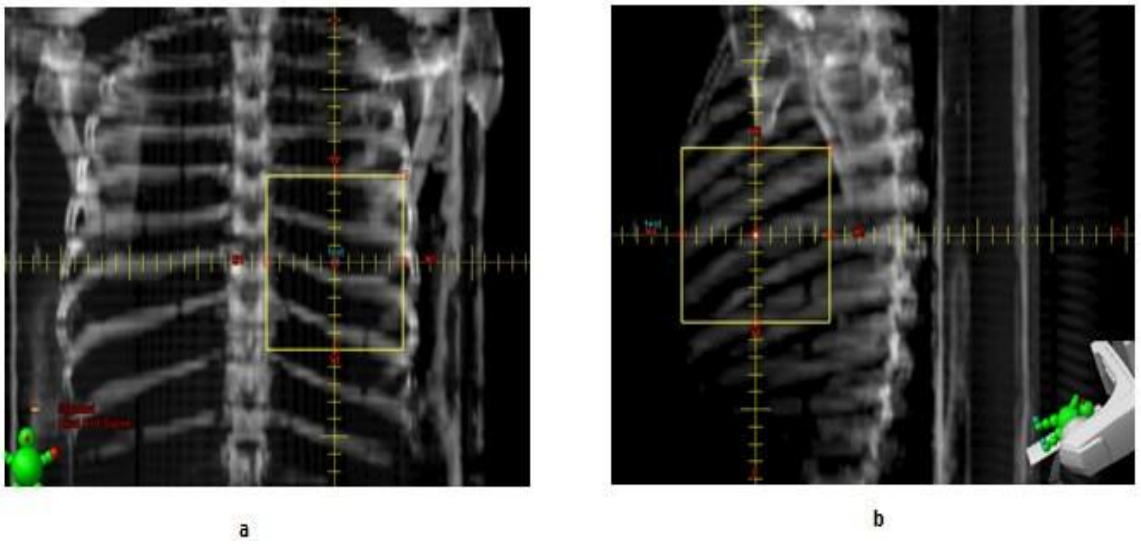

Figure. 1: The digital reconstructed radiograph (DRR) constructed by the TPS used to help us during simulation step. (a) Anterior view, (b) lateral view.

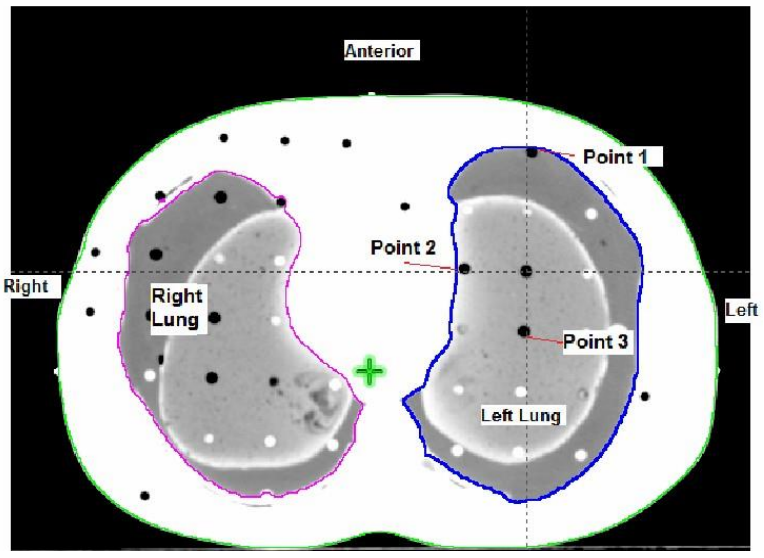

Figure. 2: CT chest cross section of RANDO phantom demonstrates the internal structures and the positions of points. 1, 2 and 3

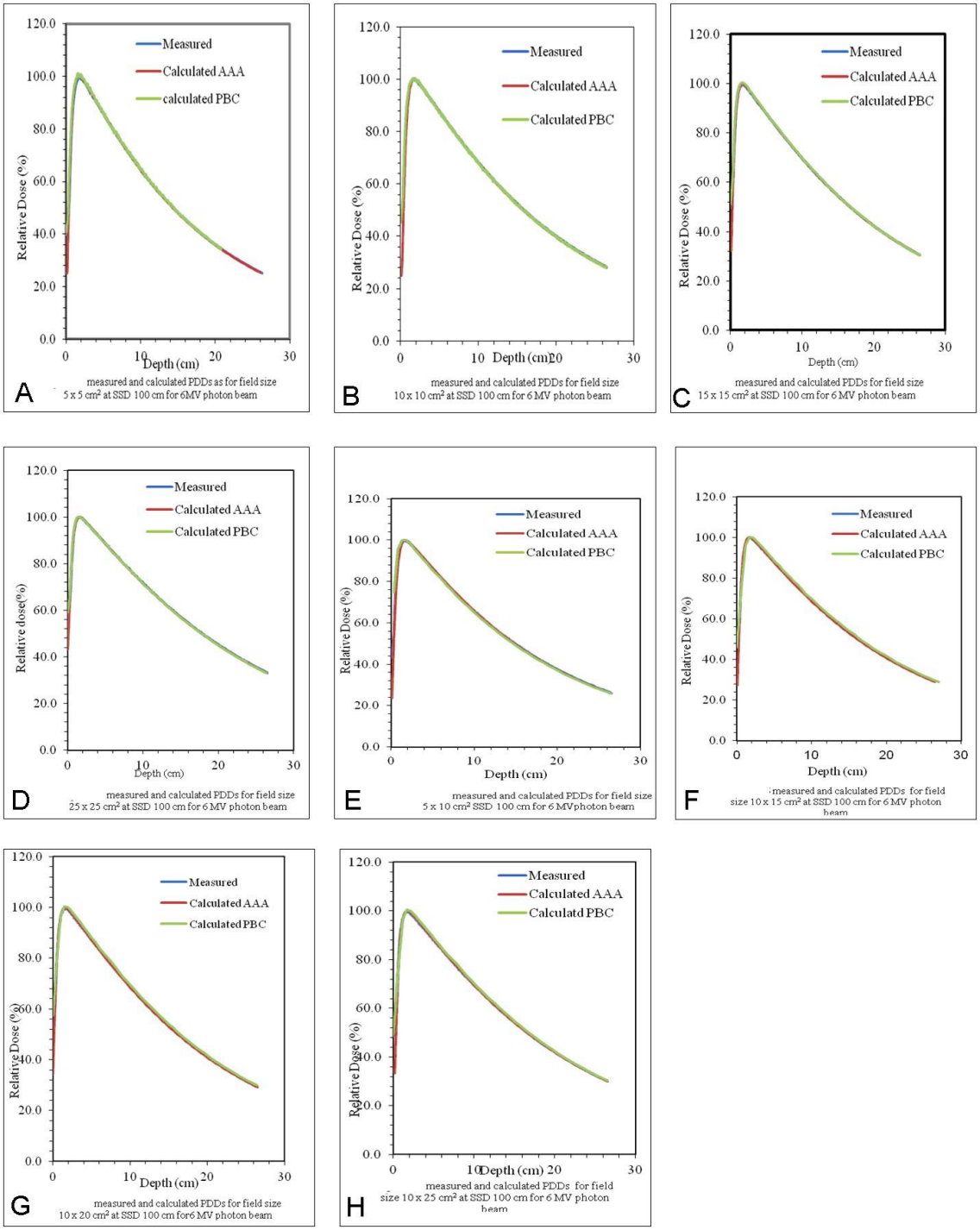

Figure. 3: Measured and calulated PPDs for different field sizes:
A: 5 X 5 cm^2
B: 10 X 10 cm^2
C: 15 X 15 cm^2
D: 25 X 25 cm^2
E: 5 X 10 cm^2
F: 10 X 15 cm^2
G: 10 X 20 cm^2
H: 10 X 20 cm^2

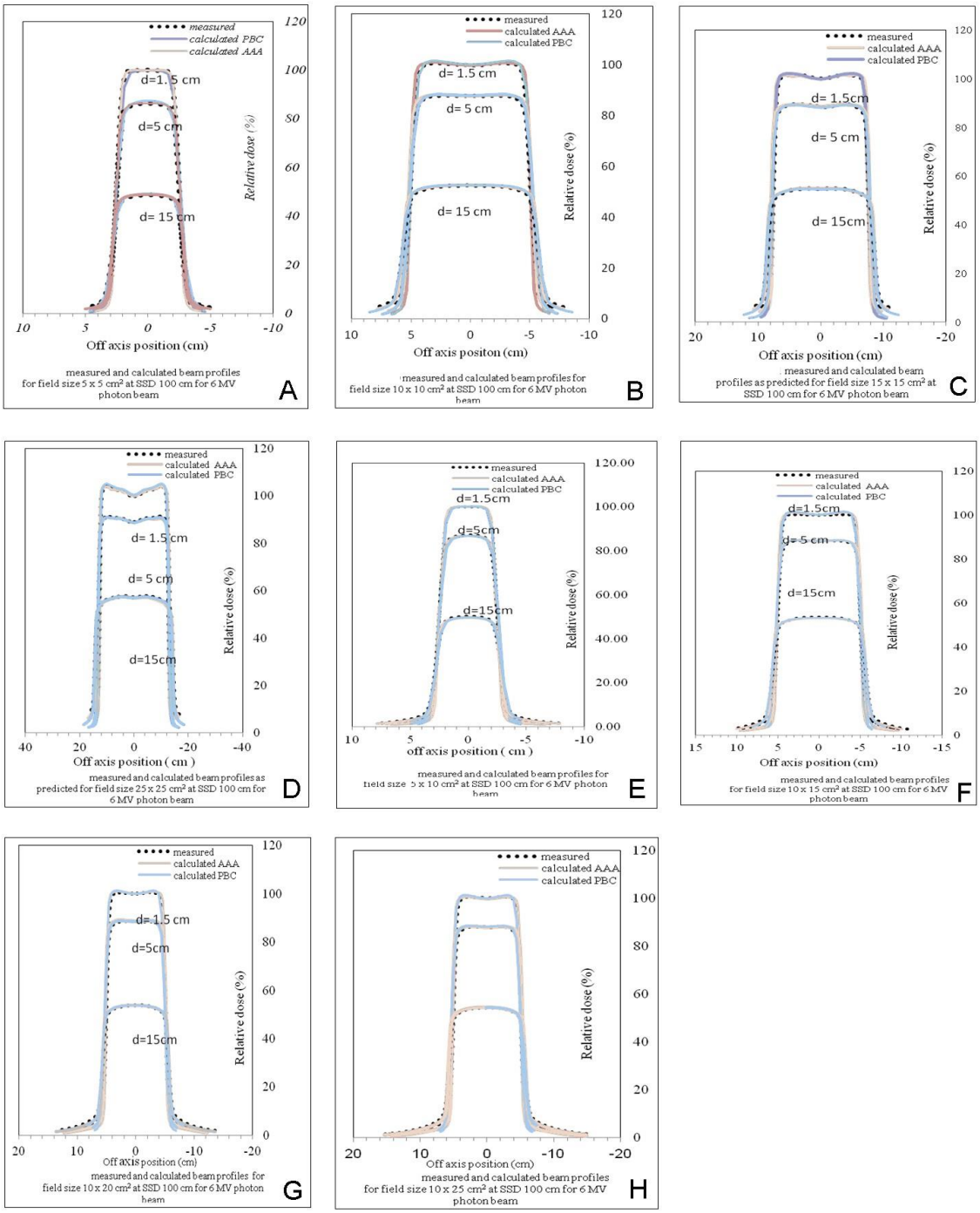

Figure. 4: Measured and calulated beam profiles for different field sizes:
A: 5 X 5 cm^2
B: 10 X 10 cm^2
C: 15 X 15 cm^2
D: 25 X 25 cm^2
E: 5 X 10 cm^2
F: 10 X 15 cm^2
G: 10 X 20 cm^2
H: 10 X 20 cm^2

Table 1: Results of measured doses (with TLD) and calculated doses (PBC and AAA).

Point no.	TLD (cGy)	Algorithm		Diff (%)	
		AAA	PBC	AAA	PBC
1	133	127.0	124.0	-4.5	-6.7
2	100	101.4	95.0	1.3	-5.2
3	93	94.5	87.7	1.6	-5.6

DISCUSSION

Since inadequate ability of TPS to correctly estimate dose distribution for the tissue heterogeneity is one of the error sources in radiotherapy process which leads to error in dose delivery to targeted PTV, so accurate modeling of primary beam attenuation and lateral scatter due to different tissue heterogeneities along beam bath in order to avoid overestimation or underestimation of the dose.

Results from water phantom measurement showed good agreement between measured PDDs and beam profiles and calculated ones; although, each of AAA and PBC has a different approach for beam modeling, they showed good agreement in homogenous medium. However, the deviations occurred in profile shoulder in PBC calculated data may be due to inaccurate accounting of scattered photons from flattening filter and jaws, while for AAA it can be referred to the insufficient ability of the algorithm to account for the electron commination especially at depth of $D_{max.}$ These results agree with those obtained by Gifford *et al*[9] depth of D_{max} Fogliata *et al*[10] and Cozzi *et al*[11].

From TLD measurements, AAA results were better than those of PBC however dose underestimation of -4.5% was observed at lung bone interface. Also, dose overestimation of 1.6% in the low density tissue (lung tissue) agrees with Robinson's evaluation of inhomogeneity correction capabilities of AAA. These results also agree with the findings of van Esch *et al*[12] that AAA improved the accuracy of dose calculations. Particular progress was made with respect to the penumbra and low dose region.

CONCLUSION

In this study we investigated the ability of AAA to account for tissue heterogeneity using rando humanoid phantom. The results that work showed that AAA made an improvement in beam modeling in homogenous medium (water phantom) measurement for both PDD and bam profile measurement over PBC, however modeling of electron contamination is not accurate enough. Also measurements in heterogeneous medium (rando phantom) showed an increased ability of AAA to model for dose distribution over PBC.

REFERENCES

3. International Commission on Radiation Units and Measurement (ICRU), (1993). Prescribing, Recording, and Reporting Photon Beam Therapy. ICRU Report 50, Betheda, Maryland, USA.
4. International Commission on Radiation Units and Measurement (ICRU), (1987). Use of computers in external beam radiotherapy procedures with high energy photons and electrons. ICRU Report 42, Betheda, Maryland, USA.
5. Breitman K, Rathee S, Newcomb C, et al. Experimental validation of the Eclipse AAA algorithm. J Appl Clin Med Phys. 2007;8:76-92.
6. Gagné IM and Zavgorodni S. Evaluation of the analytical anisotropic algorithm in an extreme water-lung interface phantom using Monte Carlo dose calculations. J Appl Clin Med Phys. 2006;8:33-46.
7. Bragg CM, Wingate K and Conway J. Clinical implications of the anisotropic analytical algorithm for IMRT treatment planning and verification. Radiother Oncol. 2008;86:(276-284.
8. Robinson D. Inhomogeneity correction and the analytic anisotropic algorithm. J Appl Clin Med Phys. 2008;9:2786.
9. da Rosa LA, Cardoso SC, Campos LT, et al. Percentage depth dose evaluation in heterogeneous media using thermoluminescent dosimetry. J Appl Clin Med Phys. 2010;11:2947.
10. Rana SB. Dose prediction accuracy of anisotropic analytical algorithm and pencil beam convolution algorithm beyond high density heterogeneity interface. South Asian J Cancer. 2013;2:26-30.
11. Gifford KA, Followill DS, Liu HH, et al. Verification of the accuracy of a photon dose-calculation algorithm. J Appl Clin Med Phys. 2002;3:26-45.
12. Fogliata A, Nicolini G, Vanetti E, et al. Dosimetric validation of the anisotropic analytical algorithm for photon dose calculation: fundamental characterization in water. Phys Med Biol. 2006;51:1421-1438.
13. Cozzi L, Nicolini G, Vanetti E, et al. Basic dosimetric verification in water of the anisotropic analytical algorithm for Varian, Elekta and Siemens linacs. Z Med Phys. 2008;18:128-135.
14. van Esch A, Trillikainen L, Pyykkonen J, et al. Testing of analytical anisotropic algorithm for photon dose calculation. Med Phys. 2006; 33:4130-4148.