Volume 13, No 4 | International Journal of Radiation Research, October 2015

Dose calculations accuracy of TiGRT treatment planning system for small IMRT beamlets in heterogeneous lung phantom

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► ***Original article***

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Revised: Aug. 2015
Accepted: Sept. 2015

Int. J. Radiat. Res., October 2015; 13(4): 345-354

DOI: 10.7508/ijrr.2015.04.008

ABSTRACT

Background: Accurate dose calculations in small beamlets and lung material have been a great challenge for most of treatment planning systems (TPS). In the current study, the dose calculation accuracy of TiGRT TPS was evaluated for small beamlets in water and lung phantom by comparison to Monte Carlo (MC) calculations. ***Materials and Methods***: The head of Siemens Oncor-impression linac was simulated for 6 and 18 MV photon beams using MCNPX MC Code. The model was validated using measured percentage depth dose and beam profiles. Then, the validated model used for dose calculations for small beamlets in water as well as lung phantoms. For treatment planning purposes, the lung phantom was scanned and imported into the TPS, and then the percentage depth dose values were obtained from plans for small fields of 1×1, 2×2, 3×3 and 4×4 cm^2 in water and lung phantom. ***Results:*** For small fields in water phantom, there was a good agreement between TPS and MC for 2×2 to 4×4 cm^2 field sizes. Nevertheless, the depth doses in lung phantom showed large discrepancies between TPS and MC calculations for points inside lung and lung-soft tissue interfaces. The TPS underestimated the lung dose up to 67% and 110% for 6 and 18 MV beams compared to MC results. ***Conclusion:*** Our findings revealed that the TiGRT TPS was not able to account for lung inhomogeities in small beamlets. Besides, the TPS calculated depth doses were not accurate enough to be used for small beamlets used in IMRT of lung region.

Keywords: *Small beamlets, lung dose calculation, Monte Carlo method, full scatter convolution, TiGRT TPS.*

INTRODUCTION

Nowadays, as the radiation therapy technology develops rapidly with utilization of newly invented hardware and algorithms to provide new modalities for radiation therapy, new treatment planning systems (TPS) are proposed and advertised commercially for their better performances and affordable prices to radiation therapy departments. On the other hand, considering the difficulties associated with acceptance testing and quality assurance of TPSs, the intrinsic uncertainty in the accuracy of new algorithms provided by new companies makes this task cumbersome and tedious for medical physicists. Another point that should be noticed here is that although some of the well-known available TPSs are equipped with Monte Carlo (MC) and convolution superposition methods for final dose calculation in IMRT planning, they employ pencil-beam (PB) algorithms in the optimization process. Moreover, in some TPSs to speed up the dose calculation process analytical methods are utilized for final dose calculations instead of more accurate methods.

There are several studies on the application of Monte Carlo calculations as a reliable and

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[DOI: 10.7508/ijrr.2015.04.008]

accurate method to evaluate the accuracy of TPS calculation for complex and intricate conditions such as dose distribution inside air and lung inhomogeneities for small photon fields [1-12]. On the other hand, different algorithms are utilized in commercial TPSs to calculate dose inside and near inhomogeneities including, air, lung and bone. Of course, it has been shown that the accuracy of their calculations varies significantly according to their beam modeling characteristics, geometry of treatment site and fields. Nevertheless, most of clinically used TPSs have provided acceptable differences relative to agreement criteria (3% in most cases) in situations frequently used for three dimensional conformal radiation therapies. However, there have been several radiotherapy cases in which the most of dose calculation algorithms have shown large differences with agreement criteria such as small fields or beamlets used for lung or thorax region radiation therapy.

In the study of Fotina *et al.* on the accuracy of new algorithm of enhanced collapsed cone algorithm verses MC method, a considerable agreement (difference less than 3%) was found for new algorithm. However, a dose underestimation of 8% was reported for organ-at-risk in IMRT plans and differences up to 5% in PTV were observed for SBRT plans [9].

We found only one published work about the full scatter convolution (FSC) algorithm in the literature. In a recent study on the accuracy of the new algorithm of FSC, it was found a difference of about 5% with measurements using a thorax phantom in a 6 MV photon beam [13]. it should be noticed that in above mentioned study, the experiment setups included the standard situations used for quality assurance of TPSs before clinical use and the performance of TPS was not verified for small fields and beamlets used in IMRT of thorax region.

In the current study, the accuracy of the FSC algorithm implemented on TiGRT TPS, a newly released system was verified versus MC calculations with MCNPX code for small beamlets ranging from 1×1 to 4×4 cm^2. A MC model of linac was built and used for dose calculations inside water and inhomogeneous lung phantoms resembling the lung irradiations.

MATERIALS AND METHODS

We used a slab phantom to obtain the required geometry for treatment planning and also for MC calculations. The schematic representation is shown in figure 1.

For lung phantom, Perspex with the thickness of 4.5 cm, 12 cm cork with density of 0.25 g/cm^3 resembling the lung and 4.5 cm Perspex under cork were used. For both TPS and MC calculations the fixed source to skin distance of 100 cm were used. The phantom was scanned with conventional X-ray CT scanner in helical mode with slice thickness of 5 mm and the images were transferred into the TiGRT planning system. Then, the depth doses on the central axis were calculated by TPS for inhomogeneous lung phantom. Also, the depth doses of the small beamlets in water phantom were calculated using the virtual water phantom module of the TPS.

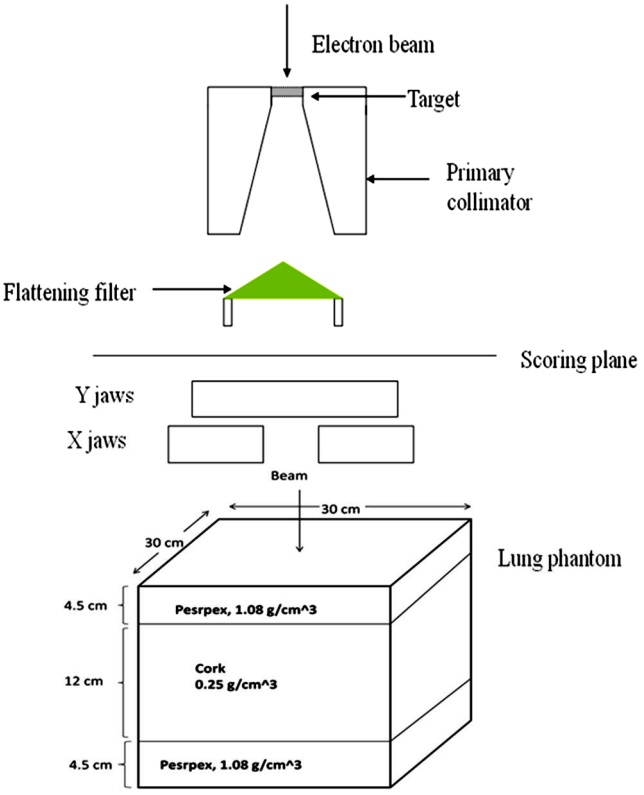


Figure 1. The simulated geometry of linac head and lung phantom.

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[DOI: 10.7508/ijrr.2015.04.008]

MC simulations

The head of Siemens Oncor-impression was simulated by MCNPX code (2.4.1) [14]. The model was consisting of electron target, primary collimator, flattening filter and secondary collimators based on the manufacturer's provided information (figure 1). The secondary collimators jaws in x-axis was multi-leaf collimator (MLC) 41 pairs and the width of leaves at the isocenter was 1 cm. However, to avoid the complexity of inter-leaf leakage problems and difficulties of the beam validation, the MLC was not simulated in our model and MLC was simulated like y-axis jaws. We assumed the uncertainty of less than 1.5% in our results for the last simplification. For MC dose calculation inside the phantoms, a phase space file (PS) of about 10 GB was generated by scoring the particles crossing a plane just above the secondary collimators for both energies. Then these PS files were used for second part of depth dose calculations, as in the second part the PS file used as a source of photons and only the opening of secondary collimator was altered to provide required field sizes for the next calculations.

For model validation, the percentage depth doses and beam profiles for 5×5 and 10×10 and 20×20 cm^2 fields were calculated by MC model and were compared with measurements in water phantom. The primary electron energy was set to 6.1 MeV and 18 MeV after its tuning by comparison of measured and calculated percentage depth dose (PDD) curve of 10×10 cm^2. It was done according to the methods used in previous papers on MC modeling of medical linacs (see reference 16 and 17 for more detailed information). The comparison of calculated PDD in water phantom for 5×5 and 10×10 cm^2 was shown in figures 2 and 3. The measured PDD curves and beam profiles of both photon beams which had been used as the basic beam data for TPS installation was used for MC model validation. Moreover, the beam profiles of the same field sizes were also compared and the model was validated for further applications. It should be noticed that the beam profile comparison were not demonstrated because of limitations in number of figures. For depth dose calculations inside water and inhomogeneous lung phantom a column of scoring cells with dimension of 2×2×2 mm^3 was defined in the central axis of beam using the Lattice card, a command in MCNPX code. The dose deposition was scored by *F8 tally which scored the deposited energy inside the cells in terms of MeV. For depth dose calculation inside the lung phantom the calculated values in terms of MeV was changed to MeV/g and the PDD was calculated.

Two types of verification were performed, first the accuracy of TPS calculations in homogenous water phantom for small fields sizes less than 5×5 cm^2 were compared with MC results. Second, the PDD curves for the same field sizes were used to evaluate TPS performance in lung phantom. The photon and electron energy cut-offs of 0.5 and 0.01 KeV was used for MC simulations. The MC runs were performed on a desktop computer and the statistical uncertainty of less than 1.5% was obtained in all MC calculations.

TiGRT treatment planning system

This system was designed by LinaTech (Sunnyvale, CA,USA) for dose calculations in external photon and electron beams. It supports all commercial medical linear accelerators with different multi-leaf collimators as well as step-and-shoot and dynamic IMRT methods. For dose calculation inside patient body it uses X-ray computed tomography images. Also, it is capable to perform fusion of other imaging modalities including MRI, SPECT and PET with X-ray CT images for efficient treatment planning. According to its user manual, TiGRT uses an exclusive algorithm named as full scatter convolution (FSC) developed by manufacturer to meet the needs for fast and accurate calculations. This algorithm uses the basic beam data collected during the commissioning of the machine including tissue-maximum ratios (TMR), beam profiles, total scatter factors and collimator factors. The dose calculation time is under ten seconds per beam for conventional and three dimensional conformal techniques. The overall accuracy of better than 3% has been reported by used manual. According to the TPS

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user manual, the FSC algorithm separated the absorbed dose D in a given point into the primary dose D_p and the scatter dose D_s:

$$D=D_p+D_s \qquad (1)$$

The primary dose $D_p(\vec{r})$ is calculated based on convolution algorithm and according to the following formula:

$$D_p(\vec{r}) = \iiint \Phi_p(\overrightarrow{r'})\, k_p\,(\vec{r}-\overrightarrow{r'})dV' \quad (2)$$

In which $\Phi_p(\overrightarrow{r'})$ denotes the photon fluence at the surface of a ray passing through surface to point $\overrightarrow{r'}$. $k_p\,(\vec{r}-\overrightarrow{r'})$ is the electron transport kernel which describes the dose distribution around the photon primar y interaction site.

This shows that the electron transport modeling has taken account by this algorithm and the electron dose deposition kernel can be scaled for inhomogeneities like bone, lung and air. And finally dV' is the differential calculation volume at the point $\overrightarrow{r'}$.

The scatter dose $D_s(\overrightarrow{r'})$ is derived from the following convolution equation

$$D_s(\vec{r})=\iiint \Phi_p(\overrightarrow{r'})\, k_s\,(\vec{r}-\overrightarrow{r'})dV' \quad (3)$$

In this algorithm the multiple scattering of photons is ignored and $k_s\,(\vec{r}-\overrightarrow{r'})$ is the first scatter fluence kernel. This kernel can be derived from electron transport kernel. For more detailed explanation, reading the user manual of TPS is recommended.

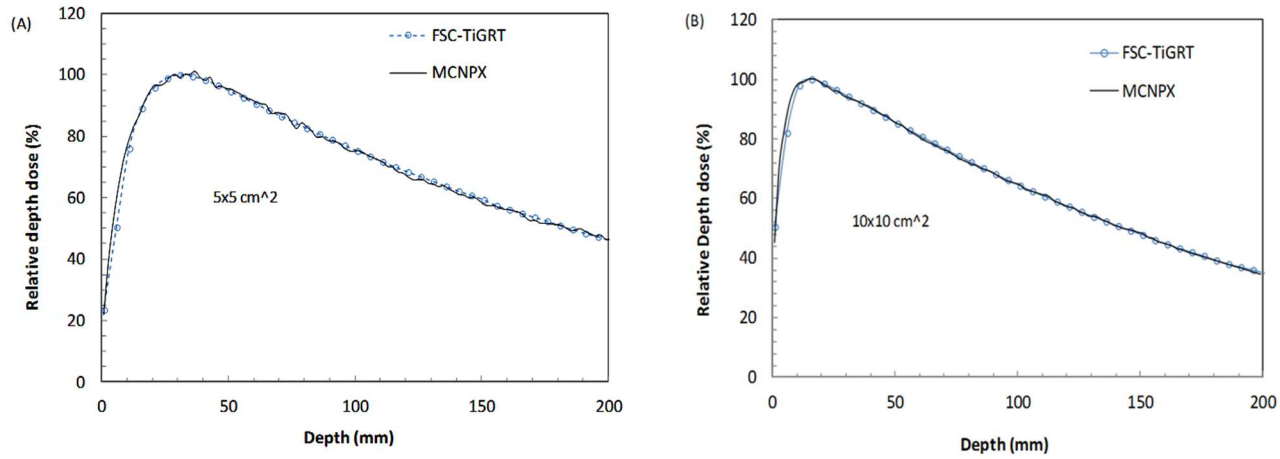


Figure 2. The comparison of relative depth doses from Monte Carlo Method and measurement in homogenous water phantom for two field sizes of (A) 5×5 and (B) 10×10 cm² for the 6 MV photon beam.

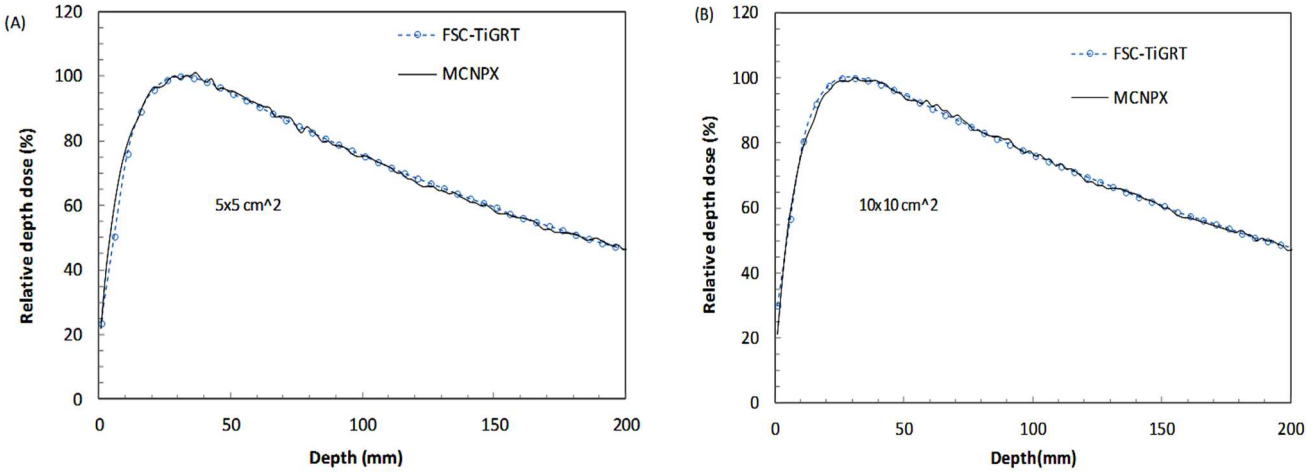


Figure 3. The comparison of relative depth doses from measurements and Monte Carlo Method in homogenous water phantom for two field sizes of (A) 5×5 and (B) 10×10 cm² for the 18 MV photon beam.

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RESULTS

The results of percentage depth dose calculations with MC model and measured data for field sizes of 5×5 and 10×10 cm^2 in the water phantom were shown in figures 2 and 3. As it can be seen that there is a close agreement between MC model and measurements. However, discrepancies up to 2% were seen between the MC model results and measurements for all field sizes from D_{max} to the depth of 20 cm. Additionally, the dose discrepancy between MC results and measurement was higher up to 10% in the buildup region for all field sizes and both energies. This can be attributed partly to the ionization chamber (IC) volume effect compared to the dose resolution of 2 mm for MC method. Besides, the dose measurement errors in build up region, where the dose gradient is steep, with IC dosimeters should be considered anyway. At last, the results showed that our MC model was accurate enough for other MC based calculation as it was intended in this research.

To evaluate the dose calculation accuracy of TPS in homogenous material like water, the depth dose for water phantom was calculated by TPS and MC for small beamlets of 1×1, 2×2, 3×3 and 4×4. It should be mentioned that TiGRT TPS uses the depth dose measurements in water phantom for field sizes from 4×4 to 40×40 cm^2 for dose calculations. Additionally, for small fields employed in IMRT beamlets the TPS applies its specific interpolation algorithm for field less than 4×4 cm^2 according to the information provided in user manual. So, the objective was to evaluate how well it calculates the depth dose for non-measured depth doses. The resulted depth dose curves in figures 4 and 5 show that the discrepancy between TPS and MC increases, when the field size reduces from 3×3 to 1×1 cm^2 for both energies of 6 and 18 MV. Also, the dose difference between TPS and MC in descending part of depth dose curve reached up to 6% for 6 MV and 8% for 18 MV photon beam. Thus, it is evident that the TPS overestimates the depth dose in water for field sizes less than 3×3 cm^2 for all curves and energies. Moreover, the

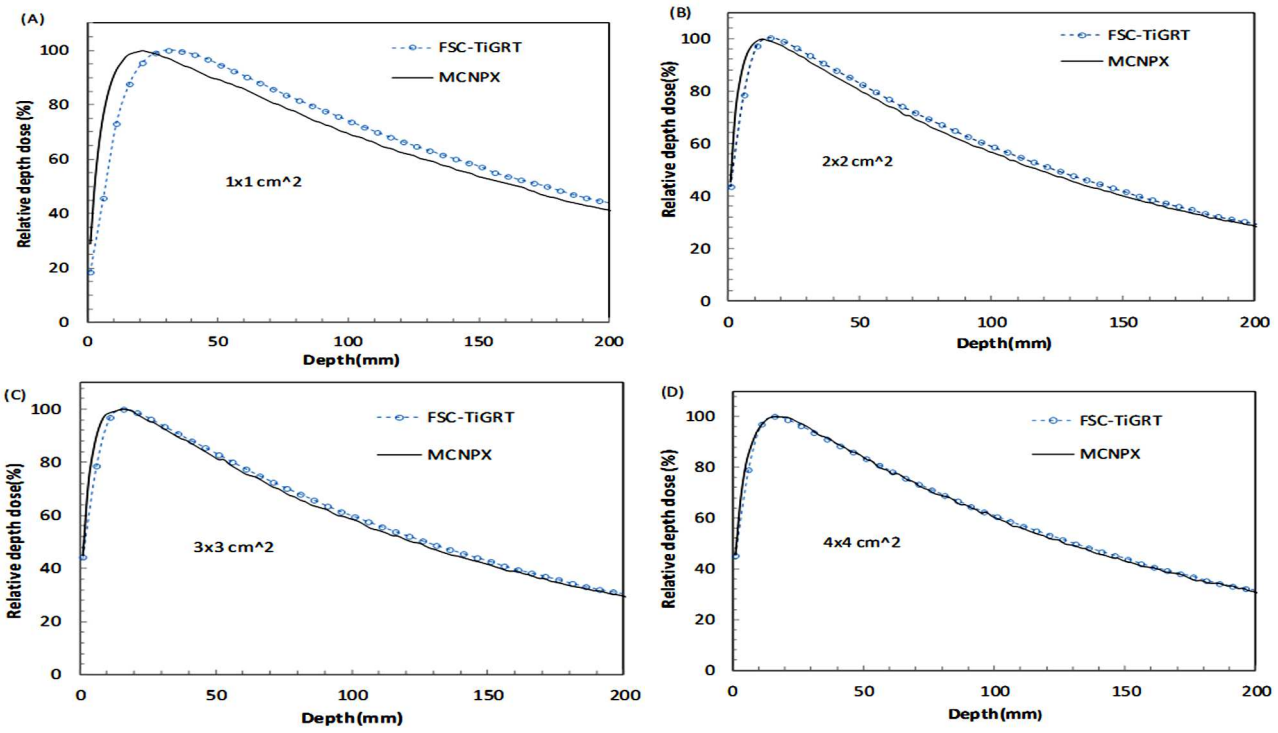


Figure 4. The comparison of relative depth doses calculated by full scatter convolution and Monte Carlo Methods in homogenous water phantom for different field sizes in the 6 MV photon beam. (A) 1×1 cm^2 (B) 2×2 cm^2 (C) 3×3 cm^2 (D) 4×4 cm^2.

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depth of D_{max} was decreased for smaller field sizes for MC calculations, while in TPS results, the depth of D_{max} was constant and showed no changes to field size for both energies. The maximum shift of almost 4 mm and 10 mm toward the surface was observed for 1×1 cm^2 field size for 6 and 18 MV photon beams respectively.

In figures 6 and 7, the depth dose calculations by TPS and MC were depicted for inhomogeneous lung phantom. The TPS depth doses were considerably higher for lung region for all small fields and both photon energies. Also, the magnitude of this overestimation for lung dose rose with field size reduction toward 1×1 cm^2. The difference between TPS and MC calculation was tabulated in table 1. The amount of overestimation varies with both field size and photon energy. The maximum differences of 67% and 110% were seen for 6 and 18 MV beams respectively in the field size of 1×1 cm^2. Additionally, the magnitude of overestimation by FSC method for 18 MV photon beam was almost two times higher than the differences seen for 6 MV beam in all field sizes.

Table 1. The difference between two methods or the magnitude of FSC overestimation compared to Monte Carlo method within inhomogeneous lung phantom.

Field size (cm^2)	6 MV	18 MV
1×1	+67%	+110%
2×2	+33%	+68%
3×3	+22%	+46%
4×4	+15%	+36%

* The dose differences were calculated at the depth of 10 cm inside the lung region using the followingequation:
Difference=(FSC-MC/MC)×100.

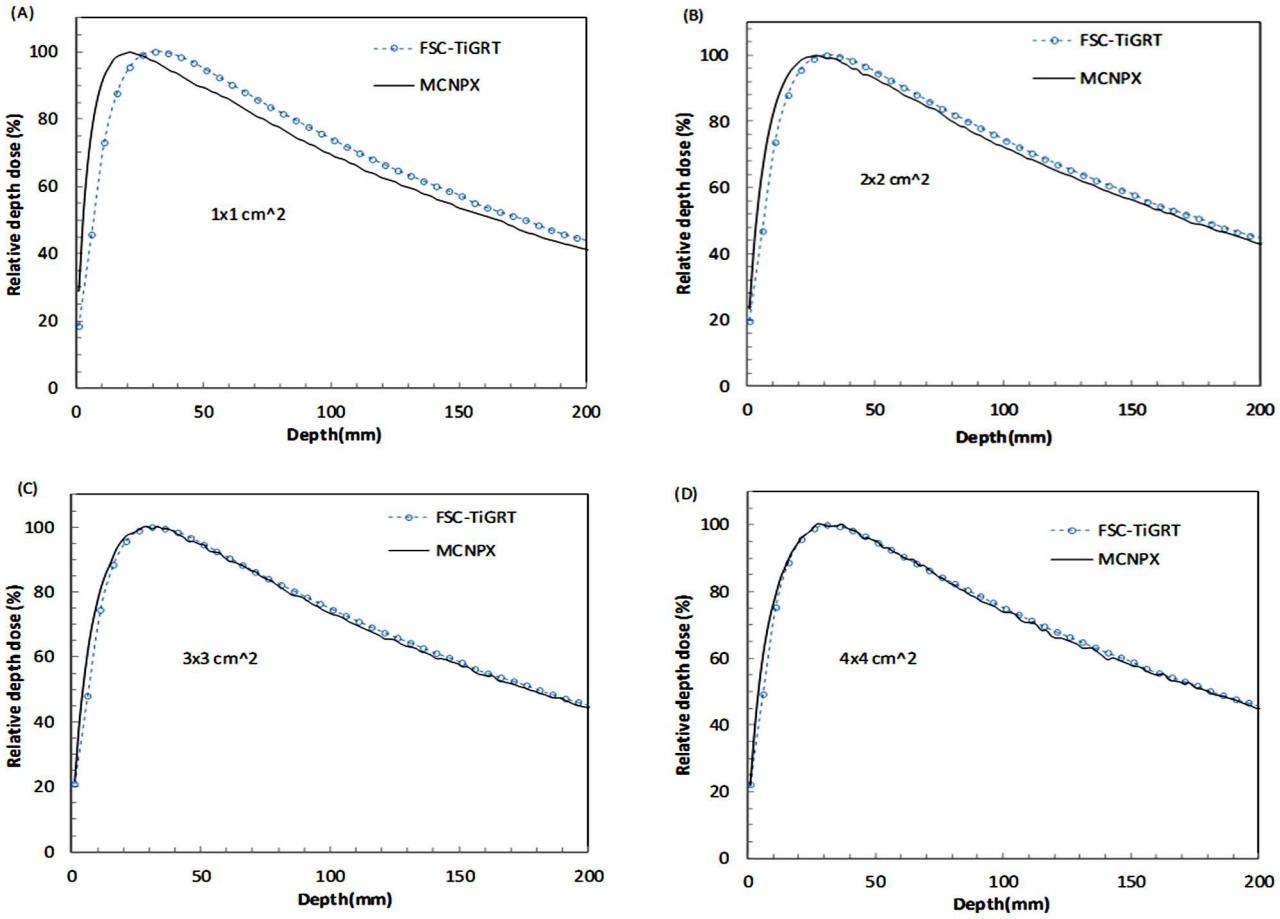


Figure 5. The comparison of relative depth doses calculated by full scatter convolution and Monte Carlo Methods in homogenous water phantom for different field sizes in the 18 MV photon beam. (A) 1×1 cm^2 (B) 2×2 cm^2 (C) 3×3 cm^2 (D)4×4 cm^2.

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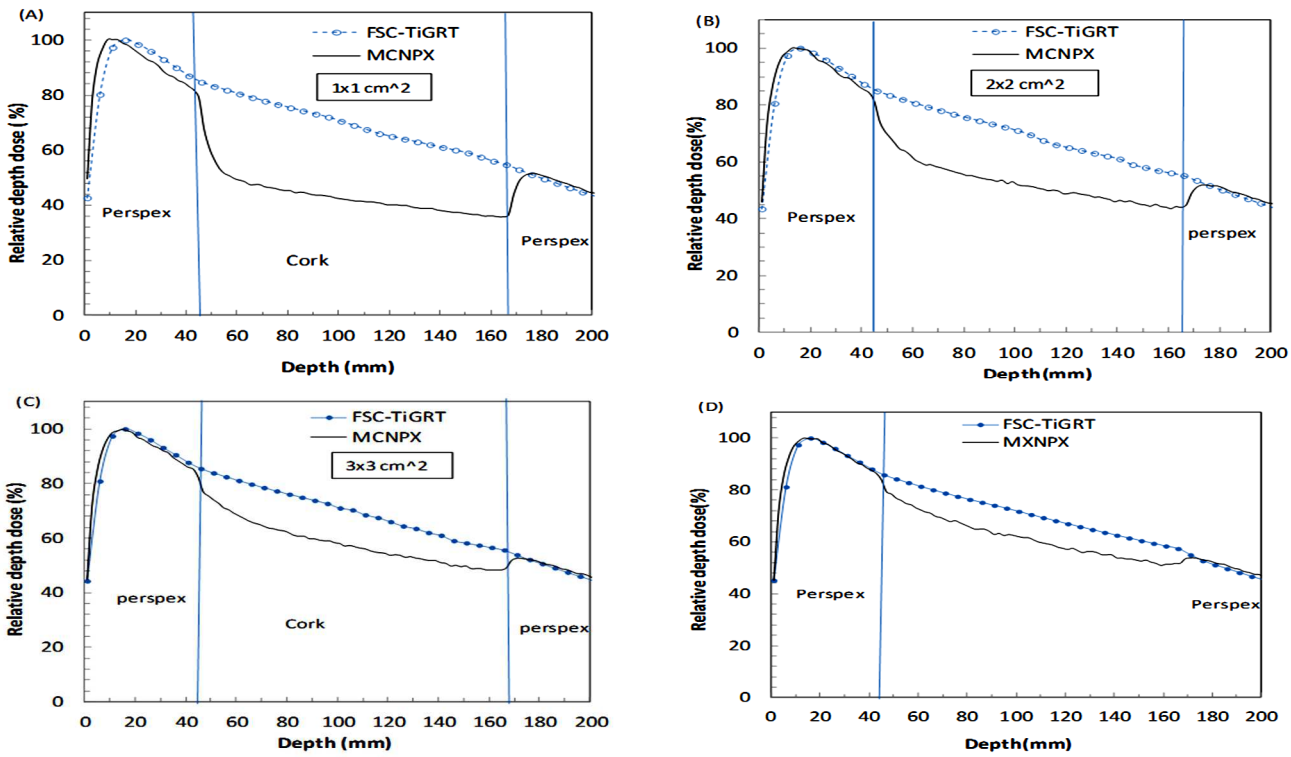


Figure 6. The comparison of relative depth doses calculated by full scatter convolution and Monte Carlo Methods in lung phantom for different field sizes in the 6 MV photon beam. (A) 1×1 cm^2 (B) 2×2 cm^2 (C) 3×3 cm^2 (D)4×4 cm^2.

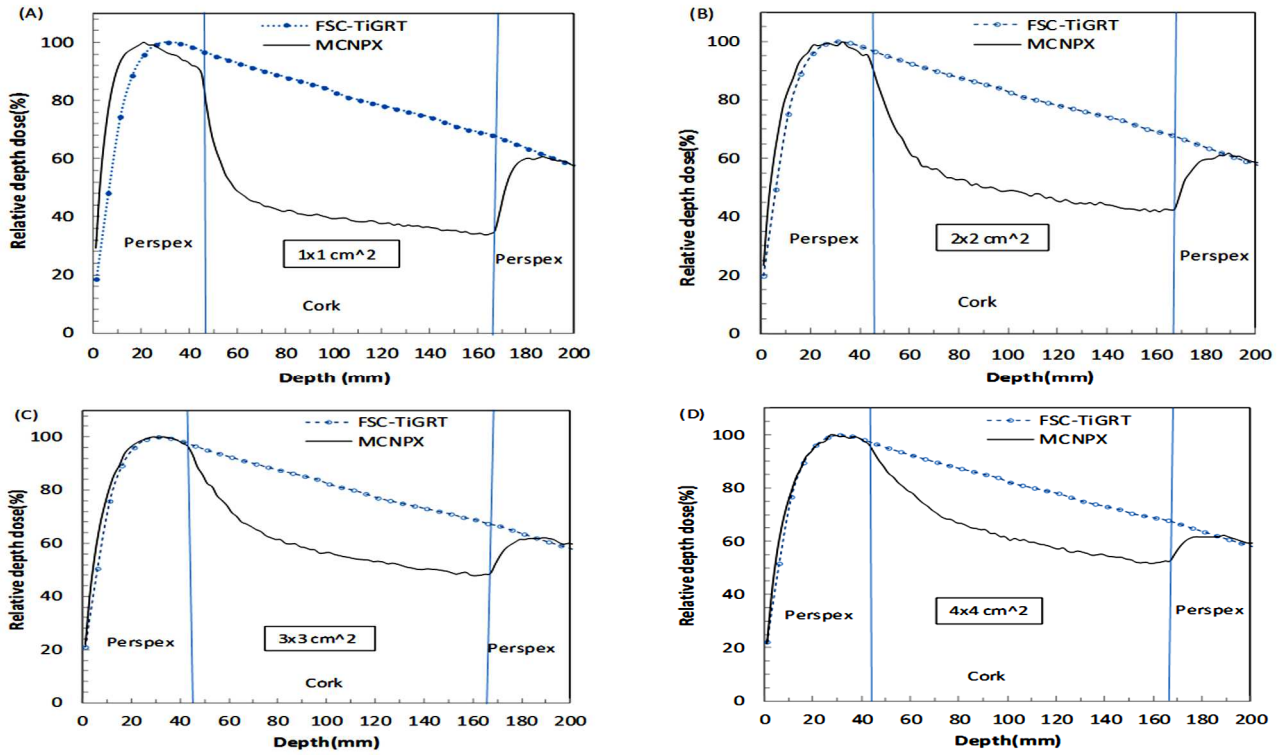


Figure 7. The comparison of relative depth doses calculated by full scatter convolution and Monte Carlo Methods in lung phantom for different field sizes in the 18 MV photon beam. (A) 1×1 cm^2 (B) 2×2 cm^2 (C) 3×3 cm^2 (D)4×4 cm^2.

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DISCUSSION

The overall figure on the performance of TPS for dose calculation inside the lung was found to be a huge overestimation for small beamlets. This arises mainly from the disability of FSC algorithm in proper modeling of electron transport inside lung where a tremendous electronic disequilibrium exists for low density material like lung in small beamlets of less than 4×4 cm^2. However, as it was mentioned earlier in method and material section, it was claimed by software developers that the electron transport was completely taken into account by FSC algorithm in TiGRT TPS. But the results of FSC was very similar to the results that had reported for other pencil beam-based algorithms applied the Batho, equivalent tissue-air ratio (ETAR) or other analytical correction methods for lung dose calculation [5,7,15-20]. In other words, FSC dose calculation results in lung were identical to other algorithms with no secondary electron transport modeling. Consequently, it can be concluded that the secondary electron transport in lung and under electronic disequilibrium is not accurately taken into account by FSC algorithm. Another point that should be noticed is the dose build up and build down at the lung- soft tissue interfaces. As it can be seen there was remarkable dose build up and build down for the studied field sizes, where the range of dose variation was larger for smaller field sizes and increased dramatically from 6 to the 18 MV beam. As it was explained by other studies, the range of secondary electrons is long inside lung, so the energy is deposited far from the primary photon interaction site and it leads to large electronic disequilibrium area at the tissue-lung interfaces. So, a sharp dose drop-off occurred at the soft tissue-lung interface in small beamlets with its maximum in 18 MV and 1×1 cm^2. Also, a steep dose build up was seen at the lung-soft tissue interface for all cases in the current study. However, it was evident from our TPS results that dose variations at interfaces was not predicted by TPS and electron transport modeling was not able to consider the complex dose deposition of electrons at interfaces.

Our results were in agreement with the results of Carrasco *et al.* on the Cadplan TPS that used ETAR method for lung dose calculation [3]. They reported a dose overestimation of 39% in lung equivalent material for a 2×2 cm^2 and 18 MV photon beam. However, in our study FSC algorithm showed an overestimation up to 68% for absorbed dose inside lung-like material for the same field size and photon energy. Finally, they showed that the collapsed cone algorithm of Helax-TMS TPS and MC simulation calculated the lung dose accurately and the results correlated with measurements with a 2% difference inside the lung.

In a study by Chen *et al.*, the accuracy of Eclipse TPS and MC-based TPS was evaluated using film dosimetry for stereotactic, single-dose irradiation of lung tumors [4]. The MC-based TPS showed a difference about 1% with measurement while the discrepancy of +15% (overestimation) for a tumor inside lung was seen for Eclipse calculations. Their study was performed on 35 clinical cases with different field sizes, and in all cases, the PB based algorithm of Eclipse overestimated the tumor dose inside lung compared to film dosimetry results. As explained by previous studies[5,7,15-20], the overestimation of lung dose or tumor inside the lung comes from the lack of electron transport in dose deposition modeling of algorithms. In a similar study, Mesbahi *et al.* assessed the performance of Eclipse TPS for dose calculation inside lung for field size of 4×4 cm^2 using measurement with a small ionization chamber and thorax phantom [18]. Two PB-based algorithm of modified Batho (MB) and ETAR were used for calculations. The dose overestimations of 33% and 28% were seen respectively for MB and ETAR methods in a 15 MV photon. In the current study the overestimation of 36% was seen for FSC algorithm for the same field size and 18 MV beam. And, it can be deduced that the performance of FSC method is very similar to MB method in terms of accuracy. However, it should be mentioned here that the MB method does not consider the electron transport in dose calculations.

In a recent study on the performance of new TPSs, the accuracy of finite-size pencil beam/ equivalent path-length (FSPB/EPL) against MC method were assessed for SBRT dose

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computations in serial tomotherapy treatments [21]. Comparing FSPB/EPL results with MC, this PB-based method overestimated minimum doses to the clinical target volume and planning target volumes by an average of 18%, and 22% respectively. It was concluded that the dose overestimation by FSPB/EPL may influence local tumor control rates, and it should be considered while the faster, but less accurate dose calculation methods are used.

To sum up, it should be mentioned that significant overestimation of FSC algorithm inside lung were seen for all studied beamlets. Besides, the results indicated that like other PB-based algorithms its inaccurate dose calculations stems from improper electron transport modeling inside low density material of lung.

CONCLUSION

The accuracy of a new commercial TPS was evaluated using the MC method as a reference method for dose calculations for small beamlets used in IMRT where the accurate dose measurement is not feasible. The results showed eventually that the implemented algorithm of FSC was not able to calculate accurately the lung dose for studied small fields. The dose build up and build down at the soft tissue-lung interfaces was not predicted by TPS for all studied cases.

According to our results and considering the strict error tolerances used for IMRT beams, the new TPS should not be utilized for IMRT treatments of lung and thorax region. For lung tumor, the overestimation of lung dose could results in insufficient dose delivery to tumor and consequently tumor non-curative irradiation. Also, the calculated dose for other organ-at-risks in IMRT plans of thorax region could lead to erroneous dose prediction and consequently compromise the treatment outcome.

ACKNOWLEDGMENTS

The authors would like to thank Tabriz University of Medical Sciences for their support. Also, the authors would like to thank radiation oncology department of Tabriz international hospital for providing the treatment planning systems for the current research.

Conflicts of interest: *none to declare.*

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