Original Article

Evaluation of the photon dose calculation accuracy in radiation therapy of malignant pleural mesothelioma

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ABSTRACT

Background: Photon dose distribution of malignant pleural mesothelioma (MPM) in matched photon-electron technique is influenced by media inhomogeneity, lateral electronic disequilibrium at interfaces and narrow field. These may influence the dose calculation accuracy, calculated by treatment planning systems (TPS). This study aimed to evaluate the dose calculation accuracy of TiGRT TPS in radiation therapy of MPM.

Materials and Methods: 18 MV photon beams of ONCOR Siemens linear accelerator was simulated using EGSnrc Monte Carlo (MC) code. Data of four patients were used to compare TPS and MC results in different regions included: Open and in-field, under shied and out of field regions.

Results: Compared to MC results, the TPS overestimated the pleura dose coverage (90% of prescribed dose) about 3–12 mm, and also it overestimated the dose in under the shielded regions of lung (4–74%). While the TPS underestimated the dose profile width about 1–16 mm in low dose region (<50% prescribed dose) as well as the out of field region dose (4–100%).

Conclusions: Results showed that TPS underestimated the dose in out of field and overestimated the dose in under the shielded regions. Unlike MC measurements, TPS calculation showed adequate pleural dose coverage. Based on the results, MC calculation can be used in matched photon-electron beam radiation therapy of MPM to modify the TPS photon dose calculations in the presence of heterogeneity, interfaces, and shield in MPM radiotherapy.

KEY WORDS: EGSnrc Monte Carlo code, inhomogeneity, malignant pleural mesothelioma, radiation therapy, TiGRT

INTRODUCTION

Malignant pleural mesothelioma (MPM) is a rare tumor which develops on the lining of the lungs called the pleural. It has been stated that asbestos is the main etiological agent of MPM.[1]

Radiotherapy can be used as an adjuvant treatment in combination with surgery for the treatment of mesothelioma. One of the main challenges in radiotherapy of MPM is a large area of target volume that involves almost the entire hemithorax.[2,3] Within and adjacent to this treatment volume, there are many normal at risk structures such as lungs, heart, kidney, liver, and spinal cord. These structures are sensitive to radiation and consequently, it is too difficult to achieve an optimized treatment plan that deliver a high sufficient dose to the target volume, while minimizing the dose to these critical organs.[4]

One of the techniques used to treat this tumor type is a technique of matched photon-electron. In this method, photon beam treatment is given through the anterior and posterior fields that encompass the entire involved hemithorax.[5,6] Protection of contralateral lung from low doses of radiation is achieved by avoiding oblique beams. In order to spare the ipsilateral lung in the photon field, a block is added for the central part of the lungs. The heart and upper abdominal organs are also blocked.[2,3] In order to compensate the missing dose in the blocked areas, the anterior and posterior chest walls dose located underneath the heart, lung, and upper abdominal blocks are boosted with an electron field.[7] This technique has

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some disadvantages such as nonuniform dose distribution in photon-electron interface and insufficient target volume dose coverage.[4] Moreover, due to low density of the lungs, primary and secondary electrons penetrate into the lungs, and this will jeopardize the protection of lung.[8]

Photon dose calculation in matched photon-electron technique is extremely challenging.[9,10] These challenges are due to the presence of inhomogeneities such as lung tissue, lung-soft tissue interfaces, as well as difficulties related to irregular and narrow shaped radiation field, and application of blocks.[11,12] These challenges may influence the dose calculation accuracy, calculated by treatment planning systems (TPS).[13-18]

To the best of our knowledge, the photon dose calculation accuracy in radiation therapy of MPM for TiGRT TPS has not been investigated. In the present study, we investigated the effects of inhomogeneities, interfaces, narrow field, and shield on photon dose distribution calculation accuracy in radiotherapy of malignant pleural for TiGRT TPS using Monte Carlo (MC) simulation.

MATERIALS AND METHODS

Linac head simulation and its validation

The EGS4nrc MC code, BEAMnrc[19] was used to simulate the 18 MV X-ray beam of a Siemens ONCOR placed at Milad hospital (Isfahan, Iran). The linac dose distribution was calculated in water phantom using the DOSXYZnrc code.[20] This linac has six main component modules including target, primary collimator and flattening filter, ionization chamber, mirror, jaws, and multi leaf collimators (MLCs). Mirror and ion chamber were not modeled because of negligible effects on photon radiation beam. Source number 19 in BEAMnrc was used to simulate a Gaussian radial intensity distribution for the incident electron beam. A circular cross-section and full width of the spectra at half maximum (FWHM) value of 0.1 cm were used in the depth dose and also dose profile simulations. A monoenergetic electron beam was used to minimize the effect of energy spread on both depth dose curves and dose profiles. The electron energy cutoff and the photon energy cutoff were set to 0.7 and 0.01 MeV, respectively, so that these values were selected based on scientific literatures.[21-23] The numbers of histories for MC calculations were 10^8 particles. The jaws were set to produce a geometrical 40 cm × 40 cm radiation field [Figure 1].

The water phantom was simulated using the DOSXYZnrc, EGS4nrc code. A water phantom with a size of 60 cm × 60 cm × 60 cm (x × y × y) and voxel size of 1 cm × 1 cm × 1 cm was used. In high dose gradient region such as penumbra and build up, the voxel size was selected 0.3 cm × 0.3 cm × 0.3 cm.

To validate our simulation data, we have compared our MC results with the corresponding measured values by ionization chamber. Dose distribution measurements were performed

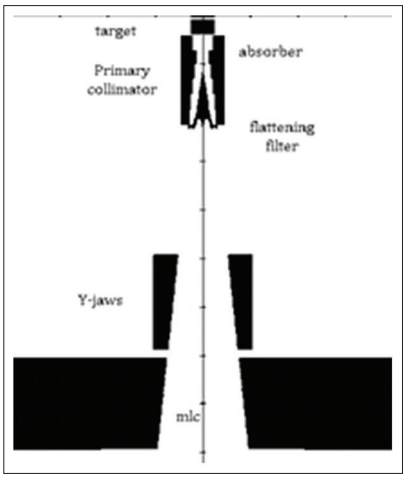

Figure 1: Components of the Siemens ONCOR linac for 18 MV photon beams and 40 cm × 40 cm field, simulated by BEAMnrc code

with photon beam energy of 18 MV, an 48 cm × 48 cm × 48 cm automatic water phantom (Medphysto mc^2, mp3, PTW, Germany) and Semiflex ionization chamber with a sensitive volume of 0.125 cm^3 (PTW, Freiburg, Germany). Measurements were performed at a source-to-surface distance of 100 cm. Percentage depth dose (PDD) and 5, 12.5, and 20 cm dose profiles were used to validate the linac head simulation.

For comparison between MC simulation and measurement results, some similar studies[21,24] use the following equation recommended by Venselaar *et al.*[25] as a criteria for the acceptance of calculation results in a water phantom:

$$\delta(\%) = \frac{D_{MC} - D_{meas}}{D_{meas}} \times 100$$

where D_{MC} and D_{meas} are the obtained dose by MC and the measured dose by ionization chamber, respectively.

δ value is different for regions with various dose values and dose gradient. Delta regions are defined as follows: δ_1 for points on the central axis beyond the depth of d_{max}, δ_2 for points in the build-up region, penumbra or interfaces of heterogeneities, δ_3 for points inside the radiation field, δ_4 for points outside the edges of the field, and δ_{50-90} (beam fringe) is the distance between the 50% and 90% point on the penumbra.

Treatment planning

In this work, computed tomography (CT) images of for four patients who previously treated with matched photon-electron technique for MPM at Milad hospital were used. The plans were created using the TiGRT version 1.2 (LinaTech, USA) TPS. TiGRT is a radiation therapy TPS for dose planning of patients undergoing external beam treatment in clinical oncology. The TPS is used to plan radiation treatments with linear accelerators and other similar radiotherapy devices with X-ray energies from 1 to 25 MV, as well as Cobalt-60, and electron energies from 1 to 25 MeV. TiGRT TPS uses a

three-dimensional (3D) photon dose calculation algorithm based on full scatter convolution.

Target volume was defined using CT images and pleura anatomical area. Critical organs in mesothelioma treatment field (right case) are ipsilateral lung, contralateral lung, liver, kidney, and spinal cord. The lead block (thickness of 7 cm) was used to shield the critical organs and also to shape the radiation field. Since TPS was not capable to select the MLC and block simultaneously, two fields in one plan was used. MLC was used to shape the first field and for second one the block was placed middle of the lung to protect the lung. Figure 2 shows the treatment planning for one of the patients.

Isocenter was selected in the middle of the field. As this point was under the block, relative dose distribution normalized to a point in the middle of planning target volume. Treatment planning was performed using 18 MV photon beam. The total dose for each patient was 5000 cGy and dose per fraction was 200 cGy.

Dose calculation in mesothelioma plan by Monte Carlo simulation and treatment planning systems

At first, TiGRT TPS was used to plan mesothelioma of patient to get the optimized dose values for radiation field and the shield for right lung (thickness of 7 cm). Then, dose profiles were calculated in selected depths. At second, tomographic phantom was simulated using CT images of patients (Siemens, Somatom Sensation 40/64) and DOSXYZnrc/ctcreate user code. Then, dose profiles were measured at the same condition of first stage. It is notable that the output of DOSXYZnrc is a kind of text file that shows the relative dose of each point in phantom (3D dose). Moreover, for each patient, number of histories was considered 100 million particles. The maximum Type A uncertainty of MC calculation was 2.46%.

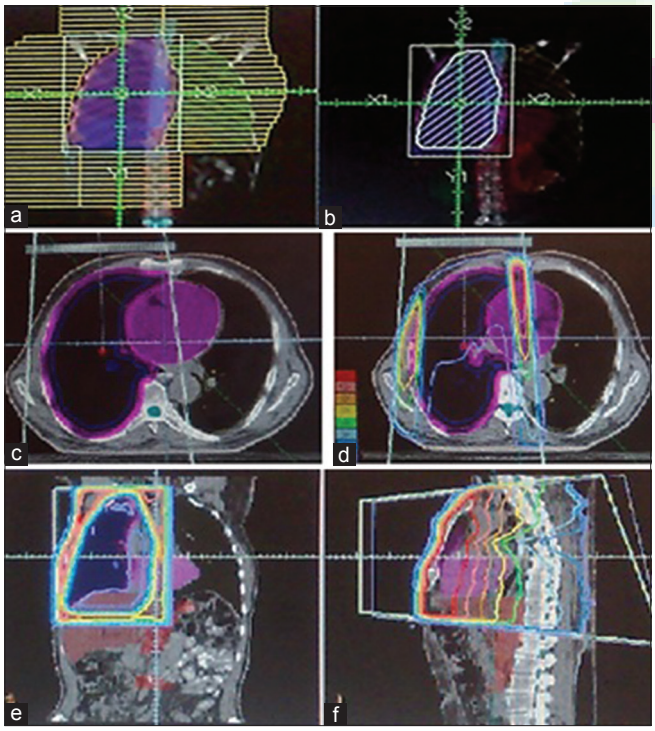

Figure 2: Treatment plan for mesothelioma case, using TiGRT treatment planning systems. Impossibility of simultaneous selection of both multi leaf collimators and block in treatment planning systems caused to plan two different fields, one for multi leaf collimators (a) and the other for block (b and c) planning target volume for mesothelioma case that cover a few millimeters in and out of the lung. (d) Dose distribution obtained by combining two fields of multi leaf collimators and block for 18 MV photon beam and the total dose of 5000 cGy in transverse plane, (e) Coronal plane, (f) Sagittal plane

Analysis of results and comparison between Monte Carlo and treatment planning systems data

To assess the dose calculation accuracy of TPS in open and in-field regions; first, dose profiles were obtained by MC and TPS at the same depth for 4 patients. Then, profile width (PW) at selected relative dose was obtained in both MC and TPS and their difference was obtained. The difference between the measured and the calculated dose is defined in following equation:

$$\delta(\%) = \frac{PW_{MC} - PW_{TPS}}{PW_{MC}} \times 100$$

To assess the dose calculation accuracy of TPS in out of field and under the shielded regions; first, MC-measured and TPS-calculated doses were determined for the same points in treatment planed of 4 patients. After that, the percent difference between calculated dose (D_{calc}) and measured dose (D_{meas}) was determined for these points. The results were analyzed to determine if there was evidence of underestimation or overestimation of does in different points by TiGRT TPS.

For analysis of the results, TRS 430[26] and TECDOC 1540[27] protocols were used. These protocols include detailed information on quality assurance of TPSs. According to these protocols, the difference between the measured and the calculated dose is defined in the following equation:

$$\delta(\%) = \frac{D_{calc} - D_{MC}}{D_{MC}} \times 100$$

Where D_{calc} and D_{MC} are the calculated dose by TPS and the obtained dose by MC, respectively.

RESULTS

Validation of the Monte Carlo simulation data of the treatment head

Validation of the MC simulation data of the treatment head was performed by comparing the simulated and the measured PDD curves and dose profiles in the water phantom. Figure 3 shows the simulated and the measured values of PDD and profiles (at depths of 5, 12.5, 20 cm) for the 40 cm × 40 cm field size and the 18 MV beam. Table 1 show difference between the measurement and the MC values for difference regions as well as to compare these differences with their tolerance limit values.

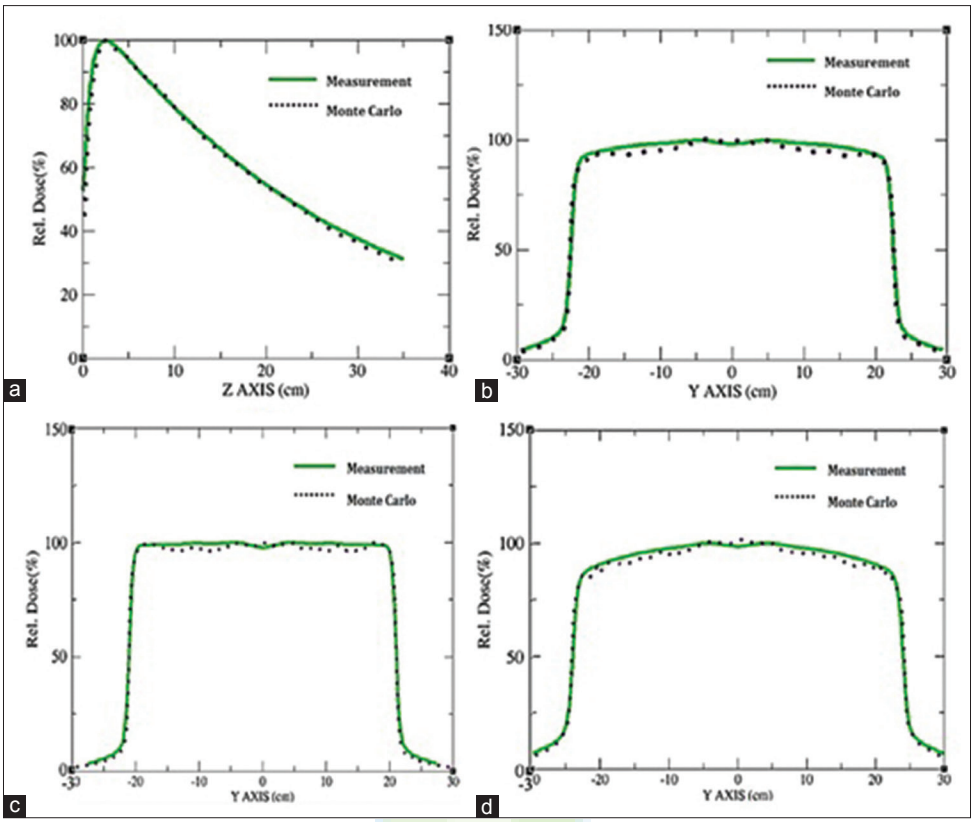

Figure 3: Percentage depth dose curve (a) and Dose profiles for depth of 12.5 cm (b), 5 cm (c) and 20 cm (d) for the 18 MV beam and field of 40 cm × 40 cm

The mean energy of the electron beam for 18 MV photons was finally set to 14 ± 0.1 MeV in MC code simulations. The FWHM was 0.1 cm for both PDD and profiles. These values were obtained by comparing the calculated and measured PDD for the 40 cm × 40 cm field size [Figure 3a].

Comparison between Monte Carlo and treatment planning systems data for mesothelioma patients

In the present study, for the first stage, dose profiles were obtained by MC simulations at selected depths for in-field, out of field, and under the shielded regions. For the second stage, dose profiles were calculated at the same depths using TiGRT TPS. Finally, differences between TPS-calculated and MC-measured doses were obtained. Figures 4-7 show the qualitative comparison between MC and TPS dose profiles for four patients in this study. Quantitative comparison of the MC and TPS results for out of the field and under the shielded regions as well as open and in-field regions were obtained in Tables 2 and 3, respectively.

Table 1: Comparison between the measured and simulated values of head linac

δ region	Difference, %	Acceptability criteria, %
δ_1	1	2
δ_2	5 (1.19 mm)	10 (2 mm)
δ_3	3	3
δ_4	20	30
$\delta_{50\text{-}90}$	0.09 mm	2 mm

Table 2: Comparison of the Monte Carlo and treatment planning system results for out of field and under the shield regions

Number of patients	Regions	TPS overestimate	TPS underestimate	Difference (%)
1	Out of field		*	4-100
	Under shield	*		4-42
2	Out of field		*	7-100
	Under shield	*		30-75
3	Out of field		*	8-100
	Under shield	*		9-21
4	Out of field		*	7-100
	Under shield	*		10-45

TPS=Treatment planning system

DISCUSSION

The presence of heterogeneous media in the treated volume influences dose distribution in MPM radiotherapy. Hence, it is necessary to apply correction factors for heterogeneities, especially for those at the lateral electronic disequilibrium at interfaces.[10-18] There are challenges to calculate MPM dose distribution, including: (1) heterogeneous media (2) interface of two medium with different densities (lung-pleura, pleura-soft tissue and lung-shield) (3) narrow, irregular field and (4) lung shield in middle of the field. These challenges may influence the dose calculation accuracy, calculated by TPS.[12-18]

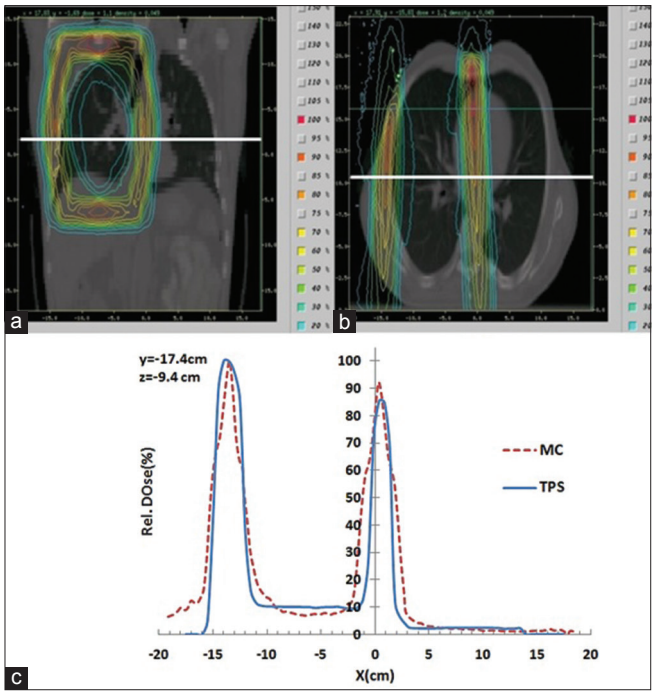

Figure 4: Mesothelioma dose distribution using the Monte Carlo in coronal plane (a) and transverse plane for case 1 (b). The white lines show location of dose profile. (c) Comparison of dose profiles between Monte Carlo and treatment planning systems in the locations, specified with white lines at (a and b)

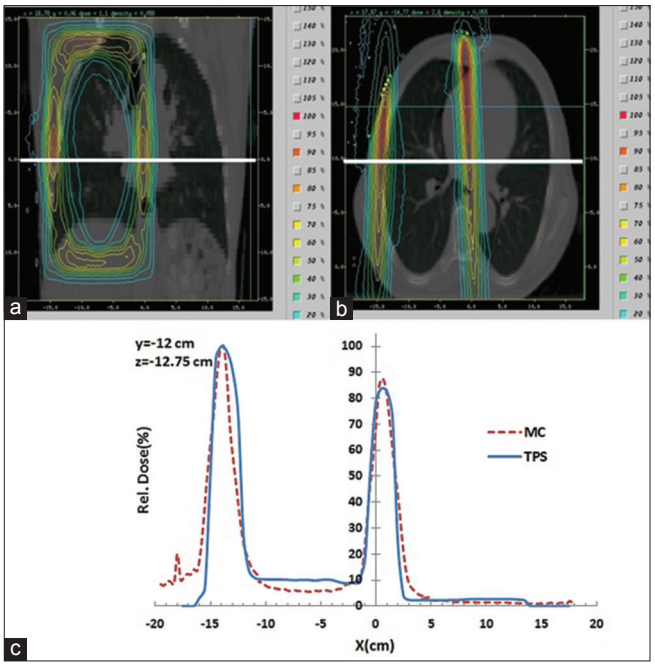

Figure 5: Mesothelioma dose distribution using the Monte Carlo in coronal plane (a) and transverse plane (b) for case 2. The white lines show location of dose profile. (c) Comparison of dose profiles between Monte Carlo and treatment planning systems in the locations, specified with white lines at (a and b)

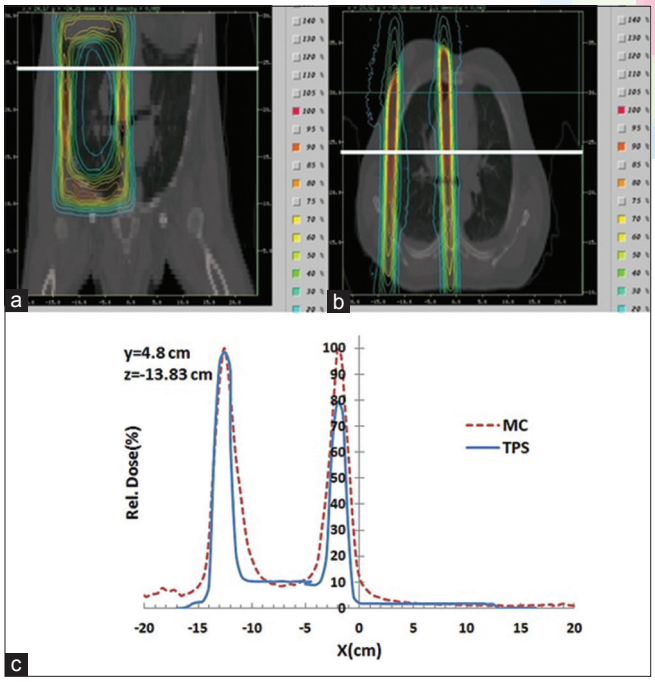

Figure 6: Mesothelioma dose distribution using the Monte Carlo in coronal plane (a) and transverse plane (b) for case 3. The white lines show location of dose profile. (c) Comparison of dose profiles between Monte Carlo and treatment planning systems in the locations, specified with white lines at (a and b)

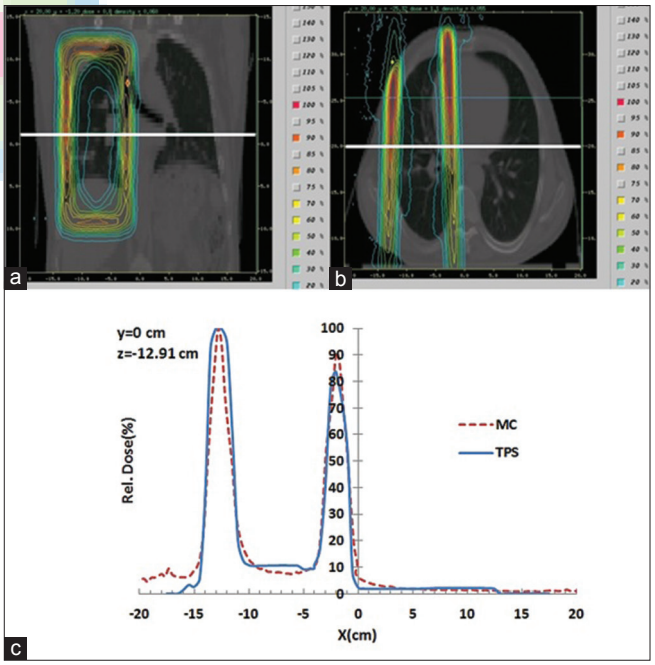

Figure 7: Mesothelioma dose distribution using the Monte Carlo in coronal plane (a) and transverse plane (b) for case 4. The white lines show location of dose profile. (c) Comparison of dose profiles between Monte Carlo and treatment planning systems in the locations, specified with white lines at (a and b)

In the present study, results showed that TiGRT TPS overestimated pleura dose coverage (90% prescribed dose) 3–12 mm (in comparison to MC). This means that TPS showed adequate dose coverage to the target (pleura), whereas MC calculation reflected inadequate dose coverage.

Table 3: Comparison of the MC and TPS results for profile width (DW) at relative dose of 20, 50 and 90% for right and left pleura

Relative dose	Case 1 (cm)			Case 2 (cm)			Case 3 (cm)			Case 4 (cm)		
	TPS	MC	Difference	TPS	MC	Difference	TPS	MC	Difference	TPS	MC	Difference
$(PW\text{-}90)_L$*	1.8	0.6	1.2	1.6	0.8	0.8	1.1	0.8	0.3	1.6	0.6	1
$(PW\text{-}50)_L$	2.8	3.1	0.3	2.75	2.55	0.2	1.75	2.35	0.6	2.45	2.15	0.3
$(PW\text{-}20)_L$	3.6	4.7	1.1	3.5	4.1	0.6	2.4	4	1.6	3.1	3.3	0.2
$(PW\text{-}50)_R$	1.9	3.1	0.2	2.3	2.2	0.1	1.4	1.7	0.3	1.9	2	0.1
$(PW\text{-}20)_R$	2.5	4.4	0.9	3	3.6	0.6	2.2	3.4	1.2	2.75	2.95	0.2

*L=left pleura, R=right pleura

Moreover, for low dose region (<50% prescribed dose), TPS underestimated the dose PW for about 1–16 mm. Comparison of TPS and MC dose profiles showed that the TPS underestimated the dose in the out of field region (4–100%). This may causes dose uncertainties in critical organs and the consequent long-term side effects. Recently, there were several studies on the assessment of dose calculation accuracy TiGRT TPS.[28,29] Bahreyni Toossi *et al*.[28] evaluated the dose calculation accuracy for outside field in breast region. They showed that TiGRT TPS compared to thermoluminescent dosimeter-measured dose generally underestimate the dose of outside points; as the mean underestimation of doses of outside field was 39%. The results of our study were consistent with the results of abovementioned study.

Other results of this study were to assess the dose calculation accuracy of TiGRT TPS in under the shielded regions. The results showed that TPS overestimated the dose (4–74%) under the shielded regions of lung.

It is notable that we used MC calculations for the assessment of dose calculation accuracy of TiGRT TPS. It has been shown that when electronic equilibrium is lost in narrow and irregular fields and due to the presence of inhomogeneities, MC calculations produce more accurate results compared to empirical and model-based algorithms.[30,31]

CONCLUSIONS

In the present study, we investigated the photon dose calculation accuracy of TiGRT TPS in radiation therapy of MPM using MC simulation. The results showed that in the presence of heterogeneity, interfaces, and shield, dose calculation accuracy of TiGRT TPS is inadequate in compared to MC simulation data. TiGRT TPS overestimated dose compared to MC measured dose for under the shielded regions, while underestimated the dose for out of field regions.

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Conflicts of interest

There are no conflicts of interest.

REFERENCES

1. van Meerbeeck JP, Scherpereel A, Surmont VF, Baas P. Malignant pleural mesothelioma: The standard of care and challenges for future management. Crit Rev Oncol Hematol 2011;78:92-111.
2. Wanebo HJ, Martini N, Melamed MR, Hilaris B, Beattie EJ Jr. Pleural mesothelioma. Cancer 1976;38:2481-8.
3. McCormack PM, Nagasaki F, Hilaris BS, Martini N. Surgical treatment of pleural mesothelioma. J Thorac Cardiovasc Surg 1982;84:834-42.
4. Baldini EH. Radiation therapy options for malignant pleural mesothelioma. Semin Thorac Cardiovasc Surg 2009;21:159-63.
5. Rice DC, Stevens CW, Correa AM, Vaporciyan AA, Tsao A, Forster KM, *et al.* Outcomes after extrapleural pneumonectomy and intensity-modulated radiation therapy for malignant pleural mesothelioma. Ann Thorac Surg 2007;84:1685-92.
6. Stevens CW, Wong PF, Rice D, Jeter M, Forster K, Zhu XR. Treatment planning system evaluation for mesothelioma IMRT. Lung Cancer 2005;49 Suppl 1:S75-81.
7. Kutcher GJ, Kestler C, Greenblatt D, Brenner H, Hilaris BS, Nori D. Technique for external beam treatment for mesothelioma. Int J Radiat Oncol Biol Phys 1987;13:1747-52.
8. Soubra M, Dunscombe PB, Hodson DI, Wong G. Physical aspects of external beam radiotherapy for the treatment of malignant pleural mesothelioma. Int J Radiat Oncol Biol Phys 1990;18:1521-7.
9. Court LE, Ching D, Schofield D, Czerminska M, Allen AM. Evaluation of the dose calculation accuracy in intensity-modulated radiation therapy for mesothelioma, focusing on low doses to the contralateral lung. J Appl Clin Med Phys 2009;10:2850.
10. White PJ, Zwicker RD, Huang DT. Comparison of dose homogeneity effects due to electron equilibrium loss in lung for 6 MV and 18 MV photons. Int J Radiat Oncol Biol Phys 1996;34:1141-6.
11. Klein EE, Chin LM, Rice RK, Mijnheer BJ. The influence of air cavities on interface doses for photon beams. Int J Radiat Oncol Biol Phys 1993;27:419-27.
12. García-Vicente F, Miñambres A, Jerez I, Modolell I, Pérez L, Torres JJ. Experimental validation tests of fast Fourier transform convolution and multigrid superposition algorithms for dose calculation in low-density media. Radiother Oncol 2003;67:239-49.
13. Hunt MA, Desobry GE, Fowble B, Coia LR. Effect of low-density lateral interfaces on soft-tissue doses. Int J Radiat Oncol Biol Phys 1997;37:475-82.
14. Epp ER, Boyer AL, Doppke KP. Underdosing of lesions resulting from lack of electronic equilibrium in upper respiratory air cavities irradiated by 10 MV X-ray beams. Int J Radiat Oncol Biol Phys 1977;2:613-9.
15. Ekstrand KE, Barnes WH. Pitfalls in the use of high energy X rays to treat tumors in the lung. Int J Radiat Oncol Biol Phys 1990;18:249-52.
16. Alam R, Ibbott GS, Pourang R, Nath R. Application of AAPM Radiation Therapy Committee Task Group 23 test package for comparison of two treatment planning systems for photon external beam radiotherapy. Med Phys 1997;24:2043-54.
17. Chow JC, Wong E, Chen JZ, Van Dyk J. Comparison of dose calculation

algorithms with Monte Carlo methods for photon arcs. Med Phys 2003;30:2686-94.
18. Jones AO, Das IJ, Jones FL Jr. A Monte Carlo study of IMRT beamlets in inhomogeneous media. Med Phys 2003;30:296-300.
19. Rogers DW, Faddegon BA, Ding GX, Ma CM, We J, Mackie TR. BEAM: A Monte Carlo code to simulate radiotherapy treatment units. Med Phys 1995;22:503-24.
20. Walters B, Kawrakow I, Rogers D. DOXYZnrc Users Manual. NRC Report PIRS; 2005.
21. Atarod M, Shokrani P, Azarnoosh A. Out-of-field beam characteristics of a 6 MV photon beam: Results of a Monte Carlo study. Appl Radiat Isot 2013;72:182-94.
22. Turian JV, Smith BD, Bernard DA, Griem KL, Chu JC. Monte Carlo calculations of output factors for clinically shaped electron fields. J Appl Clin Med Phys 2004;5:42-63.
23. Sawkey D, Faddegon BA. Simulation of large X-ray fields using independently measured source and geometry details. Med Phys 2009;36:5622-32.
24. Reis Junior JP, Salmon H, Menezes AF, Pavan GA, Rosa LA, Silva AX. Simulation of Siemens ONCOR™ Expression linear accelerator using phase space in the MCNPX code. Prog Nucl Energy 2014;70:64-70.
25. Venselaar J, Welleweerd H, Mijnheer B. Tolerances for the accuracy of photon beam dose calculations of treatment planning systems. Radiother Oncol 2001;60:191-201.
26. Andreo P, Cramb J, Fraass B, Ionescu-Farca F, Izewska J, Levin V, *et al*. Commissioning and Quality Assurance of Computerized Planning Systems for Radiation Treatment of Cancer, Technical Report Series 430. Vienna: International Atomic Energy Agency; 2004.
27. International Atomic Energy Agency. Specification and Acceptance Testing of Radiotherapy Treatment Planning Systems, TECDOC No 1540. Vienna: International Atomic Energy Agency; 2007.
28. Bahreyni Toossi MT, Soleymanifard S, Farhood B, Mohebbi S, Davenport D. Assessment of accuracy of out-of-field dose calculations by TiGRT treatment planning system in radiotherapy. J Cancer Res Ther [In press]. Available from: http://www.cancerjournal.net/preprintarticle.asp?id=176423;type=0.
29. Farhood B, Bahreyni Toossi MT, Ghorbani M, Salari E, Knaup C. Assessment the accuracy of dose calculation in build-up region for two radiotherapy treatment planning systems. J Cancer Res Ther [In press]. Available from: http://www.cancerjournal.net/preprintarticle.asp?id=176421;type=0.
30. Van Dyk J. The Modern Technology of Radiation Oncology: A Compendium for Medical Physicists and Radiation Oncologists. Madison: Medical Physics Pub Corp; 1999.
31. Khan FM. Physics of Radiation Therapy. 4th ed. Philadelphia: Lippincott Williams and Wilkins; 2010.