

Performance Characteristics of an Independent Dose Verification Program for Helical Tomotherapy

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Abstract

Helical tomotherapy with its advanced method of intensity-modulated radiation therapy delivery has been used clinically for over 20 years. The standard delivery quality assurance procedure to measure the accuracy of delivered radiation dose from each treatment plan to a phantom is time-consuming. RadCalc[®], a radiotherapy dose verification software, has released specifically for beta testing a module for tomotherapy plan dose calculations. RadCalc[®]'s accuracy for tomotherapy dose calculations was evaluated through examination of point doses in ten lung and ten prostate clinical plans. Doses calculated by the TomoHDA[™] tomotherapy treatment planning system were used as the baseline. For lung cases, RadCalc[®] overestimated point doses in the lung by an average of 13%. Doses within the spinal cord and esophagus were overestimated by 10%. Prostate plans showed better agreement, with overestimations of 6% in the prostate, bladder, and rectum. The systematic overestimation likely resulted from limitations of the pencil beam dose calculation algorithm implemented by RadCalc[®]. Limitations were more severe in areas of greater inhomogeneity and less prominent in regions of homogeneity with densities closer to 1 g/cm³. Recommendations for RadCalc[®] dose calculation algorithms and anatomical representation were provided based on the results of the study.

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INTRODUCTION

The TomoTherapy[®] Hi-ART[®] Treatment System (Accuray Inc., Sunnyvale, CA, USA) delivers intensity modulated radiation therapy (IMRT) using a 6 MV linear accelerator mounted on a gantry which guides flattening-filter-free radiation through a binary 64-leaf multileaf collimator (MLC). Radiation is delivered in a helical manner through gantry rotation and simultaneous couch movement during treatment.^[1] This provides additional degrees of freedom over conventional C-arm linear accelerators, making it possible to achieve a better three-dimensional conformation of the delivered radiation. Onboard megavoltage computed tomography (MVCT) scan is performed immediately before treatment to correct patient positioning.^[2] Due to the complex delivery of helical tomotherapy, it is imperative to perform reliable quality assurance (QA) tests to make sure that delivered dose distribution agrees well with calculated dose distribution.

The sole commercial treatment planning system (TPS) available to date for helical tomotherapy is the Hi-ART[®] treatment planning software, which was developed in

conjunction with the Hi-ART[®] treatment system.^[3] A superposition/convolution algorithm is employed by the TPS to compute dose distributions for plan preparation.

A number of studies have been conducted to verify the dose calculation model of the Hi-ART[®] TPS, including Monte Carlo (MC) simulations and other dose calculation algorithms.^[4-7] In 2008, Zhao *et al.*^[4] evaluated the Hi-ART[®] treatment planning software with MC simulations and ion chamber measurements in an anthropomorphic heterogeneous phantom. Results from a simulated lung treatment plan showed that the TPS yielded slight inaccuracies in areas of inhomogeneity and high-dose gradients, with a noticeably lower percentage of the volume passing the 3%/3 mm criterion compared to MC calculations based on BEAMnrc and DOSXYZnrc codes.^[4]

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Accuray introduction of the TomoTherapy® H Series® in recent years was accompanied by a major update to its TPS. While retaining the same dose calculation algorithm, the system hardware was upgraded from a central processing unit-based dose calculation engine to a graphic processing unit (GPU)-based engine. A study by Chen *et al.*^[8] showed that the GPU implementation offered a drastic reduction in dose calculation time with miniscule compromise in accuracy.

Despite several papers reporting on the development of software capable of providing verification doses for tomotherapy treatment plans, there are not yet any commercial software available to perform independent dose verifications.^[6,7] Thus, to verify the TPS calculated doses, the current standard procedure is to perform patient specific delivery QA (DQA). This procedure involves the calculation of the dose distribution in a phantom using the treatment plan fluence followed by dose measurements with available dosimeters to verify that dose distributions and point doses are within tolerance limits.^[3,9] However, while these methods provide good verification to the delivered dose, they are generally time-consuming and insensitive to incorrect couch replacement or selection of density calibration curve during the planning stage.^[3]

Recently, an independent monitor unit verification calculation (MUV) software developed by LifeLine Software Inc. (Austin, TX, USA), RadCalc® has released specifically for beta testing a new update (Version 6.3.3) which allows for independent point dose calculations for tomotherapy treatment plans. Previously, RadCalc® has been used as an MUV software for conventional IMRT and has yielded good results as seen in a study by Haslam *et al.*^[10] Out of the 507 treatment cases, they examined a 1.4% average discrepancy was observed between the dose calculations of a CORVUS® TPS and RadCalc®, demonstrating that RadCalc® was a viable MUV tool.

In the present work, we configured RadCalc® for the TomoHDA™ treatment unit and TPS (Version 5.1.0.4) at our center. Twenty clinical treatment plans were examined, consisting of ten lung and ten prostate cases treated at our center between 2012 and 2016. The present study aimed to examine the feasibility of using RadCalc® as an independent dose verification software and to identify any limitations of the software. More specifically, point doses were calculated by both RadCalc® and the TomoHDA™ TPS in the planning target volumes (PTVs) and organs-at-risk (OAR) of the treatment plans to compare the doses calculated by both programs.

MATERIALS AND METHODS

RadCalc® configuration

RadCalc® was installed on a Dell Precision T5500 workstation, equipped with a 2.4 GHz Intel® Xeon® dual core processor and 24 gigabytes of random-access memory. All dose calculations were performed on this workstation. RadCalc® version 6.3.3 was used in this study. The program was configured to perform dose calculations for the TomoHDA™ system at our center.

As directed by the TR-16 TomoTherapy Reference document provided by RadCalc®, a DEMODATA.zip file containing data required to setup the tomotherapy module was loaded.

Within the physics settings of the machine, the percentage depth dose, off-axis ratio, and output factor data fields were prepopulated with values directly obtained from accuracy. The TR-16 reference suggested that it was not necessary to alter these data as they are likely consistent across all TomoTherapy® units.^[8] The default values were compared with data available for the TomoHDA™ at our center and differences were found to be within 1%, thus, the prepopulated values were used for calculation purposes.

The reference machine output for the TomoHDA™ system at our center was collected independently of the TPS. Measurements were made for 6 MV 5 cm × 40 cm beam at a source to surface distance of 80 cm and a depth of 5 cm in an open water phantom with a routine Farmer TN30010 (PTW, Freiburg, Germany) ion chamber. Output values were collected at 60 s of static exposure and converted into cGy/min. The measurement was repeated five times, and the average output was 783.85 cGy/min. This value was entered into RadCalc® as a calibration for dose calculation.

The jaw width in a tomotherapy plan can be set as 1.0, 2.5, or 5.0 cm to determine the thickness of the treatment slice at isocenter.^[1] However, the jaw width and actual width of radiation delivery is offset by a certain amount due to a finite radiation source size.^[11] The radiation and jaw offset settings for each jaw opening size were changed to match our TomoHDA™ unit.

Tomotherapy treatment planning system

TomoHDA™ System Planning Station Version 5.1.0.4 (Accuray Inc., Sunnyvale, CA, USA) was used to generate TPS dosages for comparison with RadCalc® as well as the RP (plan) and RS (structure) files required for calculation by RadCalc®. All computer tomography (CT) studies were taken with 3 mm slice thickness and the “fine” 2.4 mm × 2.4 mm × 3 mm calculation grid for tomotherapy planning. In this study, twenty patient plans were examined: ten lung and ten prostate cancer cases. The choice of treatment sites allowed us to examine the performance of RadCalc® in areas of higher inhomogeneity as well as areas of low inhomogeneity. In addition, the plans vary in jaw width settings: dynamically changing jaw widths or a static jaw width. An equal number of plans with either setting were selected to mitigate the potential discrepancies between dynamic and static jaw settings. Based on the position of the jaw at any given point, output factors for dynamic jaw plans were calculated through interpolation of available output factor data for the 1.0, 2.5, or 5.0 cm jaw widths.

Several modifications were made to the TPS plans before exporting the files for RadCalc® calculations. In helical tomotherapy procedures, the couch is often in the treatment path and causes attenuation to the delivered radiation.^[4] As a result, it is necessary to include the couch during treatment planning. In the TomoHDA™ TPS, this is automatically taken

into account since kilovolt computed tomography (kVCT) images, and corresponding image value to density table (IVDT) calibration curves are used during dose calculation. However, CT data input is not supported in RadCalc®—the density and geometry of the couch and other structures have to be manually specified. All regions of interest (ROI) needed to be properly contoured, including the couch. Two simplified contours were created to represent the upper and lower sections of the couch with differing densities as shown in Figure 1.

RadCalc® computes point doses at calculation points assigned by users and compares it to corresponding point doses calculated by the TPS. To do so, the TPS calculated doses must be inputted manually. Small (3 mm diameter) circular contours were created in the TPS to “house” calculation points in RadCalc®.^[8] Two sets of contours were inserted into the PTV [Figure 2a] and one set in each of the OARs [Figure 2b]. Identical contours were drawn at the same position for three consecutive transverse slices to form a small cylindrical volume. This allowed for the calculation of average dose within the encased volume, which was compared with the dose calculated by RadCalc®.

Finally, the dose distribution was re-optimized by the TPS, and the final dose was calculated after necessary contours have been added. All added contours were set to be ignored during the optimization procedure. The final doses were compared against those of the original treatment plan to ensure consistency before the modified plans and structure files were exported through digital imaging and communications in medicine (DICOM).

RadCalc® dose calculation

RT (plan) and RS (structure) files were imported into RadCalc®. The imported tomotherapy calculation was accessed through the conventional RadCalc® patient database.

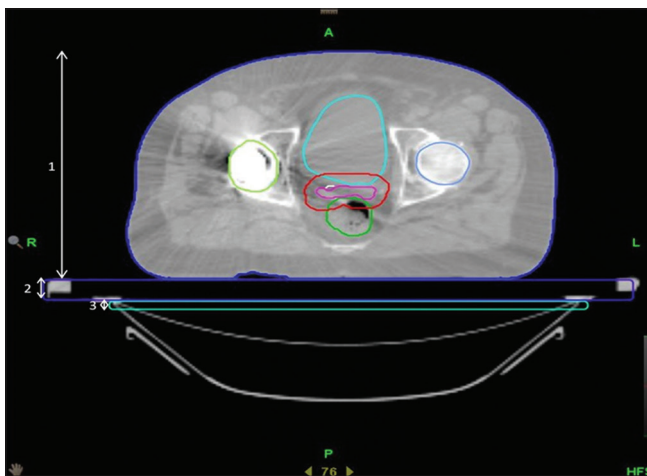


Figure 1: Transverse view of a prostate case in the TomoHDA™ planning system, before insertion of the tomotherapy couch. (1) patient anatomy with regions of interests contoured, (2) added contour for upper section of the tomotherapy couch representing a 0.05 g/cm³ density, and (3) added contour for lower section of the tomotherapy couch representing a 0.85 g/cm³ density

RadCalc®’s tomotherapy module uses a variation of the ray tracing dose calculation algorithm. The imported DICOM plan files contain information for the treatment plan, expressed as a set of control points, each representing a ray projection with a unique gantry angle, couch position, and opening time for MLC leaves (sinogram data). RadCalc® segments each ray projection into a set of beamlets and performs ray tracing to determine the dosage it delivers to a certain point. The methodology of segmentation and ray tracing was developed in a study conducted by Thomas *et al.*^[7] When a calculation point is specified, contributions of dose by each beamlet is calculated and combined to find the total dose delivered by that single ray projection. The dose contributions of each ray projection are then summated to yield the total dose delivered to the calculation point throughout the course of treatment. The number of control points were recorded and averaged between the lung and prostate cases to examine differences in the complexities of each treatment site.

Figure 3 shows the interface for RadCalc® tomotherapy calculations. Several parameters were transcribed over from the treatment planning station.

Densities of each contoured structure were entered into RadCalc. These values were determined based on taking the average of point densities in the planning kVCT images and using mass density data from the International Commission on Radiological Unit report 46 as a reference.^[12] The skin contour was set to be the external contour.

Calculation points were added by changing the view mode into two-dimensional and scrolling to the middle slice of the previously inserted circular contours as shown in Figure 4. Calculation points were placed at the center of the contour. Doses at each point were extracted from the TPS plan report and entered into the “calculation points” window.

RadCalc® is capable of taking into account the inhomogeneity of the tissue in the body. This is done through considering the density changes that occur between contours in the radiated volume and scaling the depth and effective field size correspondingly.^[13] A detailed description of the process can be found in the RadCalc® Manual.^[13] The effectiveness of the inhomogeneity correction methods was evaluated by calculating each plan using the two available correction

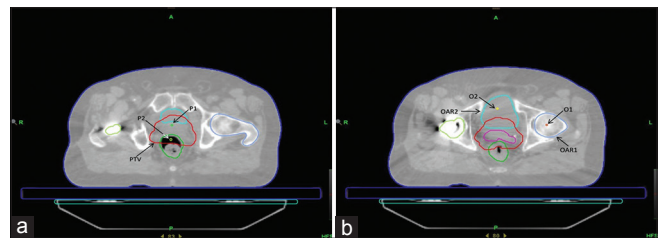


Figure 2: (a) Transverse view of a prostate case with planning target volume contours displayed. P1 and P2 are small circular contours inserted within the planning target volume, (b) transverse view of a prostate case with organs-at-risk contours displayed. O1 and O2 are small circular contours inserted into the left femur (organs-at-risk 1) and bladder (organs-at-risk 2), respectively

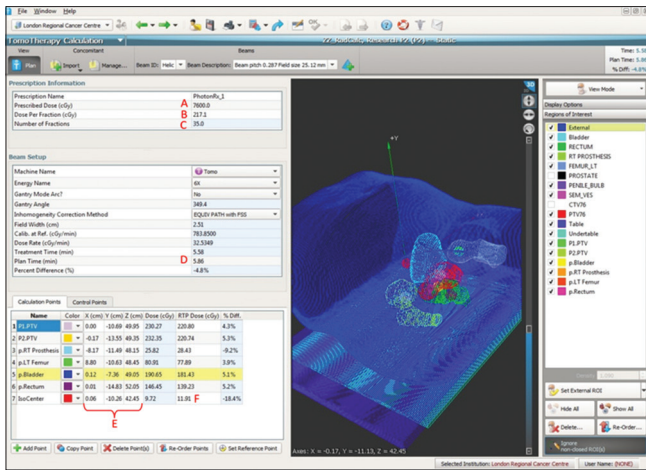


Figure 3: Tomotherapy calculation interface in RadCalc® with a prostate cancer patient loaded. Parameters include: (A) total treatment dose in cGy, (B) dose per fraction in cGy, (C) number of fractions, (D) planned time, (E) coordinates of isocenter, and (F) Radiotherapy treatment planning dose of isocenter

settings: EQUIV PATH (equivalent path length) and EQUIV PATH WITH FSS (equivalent path length with field size scaling). A third calculation was also made for reference, using no correction methods. This setting configures RadCalc® to calculate doses with all densities overridden to be 1 g/cm³ and skips any implementation of the effective depths or field sizes. Since the “none” inhomogeneity correction method was used for reference only, average value was calculated using the results of two inhomogeneity correction methods.

The “compute all beams” function was used to calculate the point doses at each calculation point. The computation time was measured and analyzed according to the number of control points and calculation points present. For each calculation point, RadCalc® outputted the calculated dose, along with the percentage error compared with the TPS dosage previously entered.

Equation (1) was used to represent the difference between the doses calculated by each program.

Dose difference

$$\text{Dose difference} = \frac{\text{RadCalc dose (cGy)} - \text{TomoHDA}^{\text{TM}} \text{ TPS dose (cGy)}}{\text{Prescribed dose (cGy)}} \times 100\% \quad (1)$$

By normalizing the difference as a percentage of the prescribed dose to the target, we were able to report the results for each calculation point relative to a common value. This was preferred over reporting local percentage differences since the percentage difference relative to the prescription dose is generally more relevant clinically.

RESULTS

Lung cases

On an average, lung treatment plans consisted of 837.3 (standard deviation [SD] = 251.2) control points. Dose calculation times

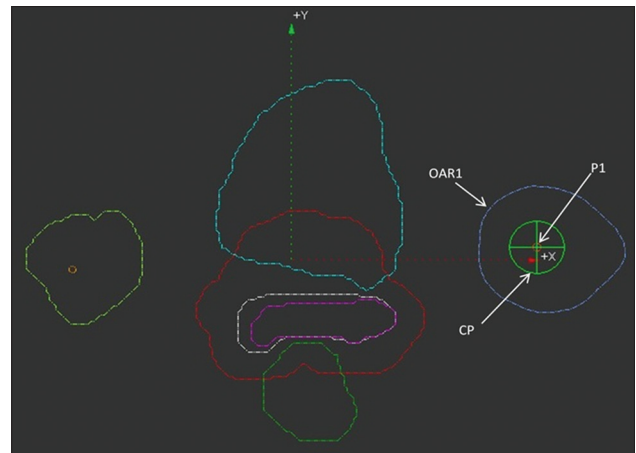


Figure 4: Two-dimensional transverse view of a prostate case in RadCalc®. P1 is a circular contour within the left femur (organs-at-risk 1). The calculation point control point was created and placed in the center of P1

using each inhomogeneity correction setting were similar. Normalized to 1000 control points, an average of 5.1 s with an SD of 0.34 s was required to compute the dose for one calculation point.

Table 1 shows the percentage differences in doses computed by RadCalc® and by the TomoHDA™ TPS, calculated using Equation (1). Points placed in the PTV region near or within lung overestimated the dose by an average of 14.35% (SD = 3.51), while points within the non-PTV region of the lung showed overestimations averaging 11.68% (SD = 13.60). In the spinal cord and esophagus, differences of 9.99% (SD = 10.92) and 10.21% (SD = 8.58) were observed, respectively. The heart exhibited a dose difference of 4.40% (SD = 5.51).

Prostate cases

A typical plan for prostate cancer patients had the following parameters on average; prostate treatment plans consisted of 674.6 (SD = 215.1) control points. Dose calculation times using each inhomogeneity correction setting were observed to be similar. Normalized to 1000 control points, an average of 4.4 s with an SD of 0.4 s was required to compute the dose for one calculation point.

Several of the prostate plans examined in this study were for patients who had hip prosthetic implants, resulting in certain femur contours being missing from the treatment plans. The presence of implants with stems made of porous coated titanium cups made of reflection titanium shell with ultra-high molecular weight polyethylene and cobalt-chromium heads were taken into account for the tomotherapy dose calculations by extending the IVDT calibration range. Although calculation points were inserted into the prostheses, they were excluded from the average to maintain the consistency of the OAR structures being analyzed. In addition, it was observed that several calculation points within or near high-density ROIs resulted in a RadCalc outputting a zero dose value. These calculation points were also excluded from the analysis and examined separately.

Table 2 shows the percentage differences in doses calculated by RadCalc® and the TPS doses for each calculation point using Equation (1). Points within the prostate PTV showed better agreement in comparison with lung cases, with an average percent difference of 5.99% (SD = 2.41). Calculations in the bladder and rectum overestimated the doses by 6.00% (SD = 1.93) and 6.41% (SD = 3.99), respectively. Slight discrepancies were observed between the two femurs cases with a percent difference of 1.56% (SD = 4.21) in the right femur and 4.60% (SD = 4.98) in the left femur.

DISCUSSION

In this study, the viability of RadCalc’s tomotherapy dose calculation module was assessed. Differences between RadCalc® doses and reference TPS doses were calculated as a percentage of the planned dose. A general overestimation of doses was observed regardless of the inhomogeneity correction method selected, and both correction methods yielded similar dose values. RadCalc®’s dose calculation employs a pencil

beam (PB) model. While the dose calculation speed of the PB algorithm is faster than other dose calculation algorithms, its accuracy is compromised.^[14] PB inhomogeneity correction algorithms are unable to effectively take into account for photons scatter and electron transport, causing inaccuracy in calculated dose values.^[15] In low-density heterogeneous tissue, such as the lung, the inaccuracies are more prominent,^[14] contributing to the high dose differences for calculation points placed with the lungs. Similar overestimations were also reported in studies which utilized the PB algorithm to compute doses in simulated lung tissue.^[16,17]

RadCalc® does not support the acquisition of patient anatomy data from kVCT images. To account for the various structures and densities of tissues and organs in the body, a structure file containing planning contours was imported. However, not all the structures were contoured in the planning stage, which can cause a misrepresentation of the patient anatomy. The high-density vertebra surrounding the spinal cord was not contoured in RadCalc®, and thus not considered during dose

Table 1: Percentage difference between RadCalc® and treatment planning system dose at calculation points averaged for 10 lung cases

| Calculation point | Inhomogeneity correction method | | | | | |
|-------------------|---------------------------------|--------|------------------------|--------|---------------------------------|--------|
| | None | | Equivalent path length | | Equivalent path length with FSS | |
| | Mean (%) | SD (%) | Mean (%) | SD (%) | Mean (%) | SD (%) |
| PTV | | | | | | |
| Outside lung | 2.00 | 5.89 | 14.16 | 3.39 | 13.80 | 3.27 |
| Within lung | -3.85 | 5.06 | 16.07 | 3.95 | 13.36 | 3.41 |
| Spinal cord | 7.28 | 8.92 | 10.03 | 10.90 | 9.95 | 10.94 |
| Right lung | 3.07 | 12.78 | 16.58 | 14.94 | 12.21 | 13.51 |
| Left lung | -4.49 | 13.79 | 11.06 | 13.15 | 6.88 | 12.70 |
| Heart | 1.90 | 3.50 | 4.41 | 5.53 | 4.39 | 5.49 |
| Esophagus | -1.00 | 7.28 | 10.28 | 8.62 | 10.14 | 8.54 |

Percentage differences in doses calculated by RadCalc® and the TPS doses for each calculation point, relative to the total dose per fraction. Calculations were performed at each calculation point placed within the PTV and OARs, averaged over ten lung cases. FSS: Field Size Scaling, SD: Standard deviation, PTV: Planning target volume, OARs: Organs-at-risks, TPS: Treatment planning system

Table 2: Percentage difference between RadCalc® and treatment planning system dose at calculation points averaged for 10 prostate cases

| Calculation point | Inhomogeneity correction method | | | | | |
|-------------------|---------------------------------|--------|------------------------|--------|---------------------------------|--------|
| | None | | Equivalent path length | | Equivalent path length with FSS | |
| | Mean (%) | SD (%) | Mean (%) | SD (%) | Mean (%) | SD (%) |
| PTV | | | | | | |
| Center | 6.27 | 2.26 | 6.46 | 2.36 | 6.49 | 2.44 |
| Off-center | 5.15 | 2.87 | 5.52 | 2.38 | 5.50 | 2.47 |
| Right femur | 0.29 | 4.27 | -1.73 | 4.17 | -1.38 | 4.24 |
| Left femur | 7.18 | 5.36 | 4.74 | 4.69 | 4.45 | 5.25 |
| Bladder | 5.86 | 2.25 | 5.97 | 1.90 | 6.03 | 1.93 |
| Rectum | 4.16 | 3.31 | 6.44 | 3.99 | 6.37 | 3.98 |

Percentage differences in doses calculated by RadCalc® and the TPS doses for each calculation point, relative to the total dose per fraction. Calculations were performed at each calculation point placed within the PTV and OARs, averaged over ten prostate cases. FSS: Field size scaling, TPS: Treatment planning system, SD: Standard deviation, PTV: Planning target volume, OARs: Organs-at-risks

computation. While the TPS would consider attenuation caused by the higher density bone tissue using CT data, RadCalc® would overestimate the dose in the spinal cord, assuming a 1 g/cm^3 density in the region.

Furthermore, densities of the contoured structures are assumed to be uniform within each contour, leading to inaccuracy especially when dealing with inhomogeneous tissue. This resulted in the high percentage dose difference for lung calculation points. Although a single density value was assigned to the lung ROI, observed densities varied greatly within the lung region. On the contrary, lower dose differences were observed for calculation points placed in the heart, prostate PTV, bladder, and rectum, which possessed high tissue homogeneity close to a density of 1 g/cm^3 . Note that since the tissue properties of these structures were similar to those assumed by RadCalc® when no inhomogeneity correction method was selected (uniform 1 g/cm^3 density), similar calculated doses were obtained whether or not inhomogeneity correction was applied.

Several prostate cases used in this study came from patients who had prosthetic hip implants. These implants were contoured in the plans and had high densities. Certain calculation points in two of these plans were excluded from the analysis due to a zero dose value being calculated when an inhomogeneity correction method was applied. Non-zero results were observed when no inhomogeneity correction method was used or when a density of 1 g/cm^3 was assigned to the prostheses. It was observed that the high densities of the prostheses in these plans, specifically 2.95 and 6.7 g/cm^3 , caused the dose in the calculation point to diminish. This underestimation of dose may be due to the inability of RadCalc®'s PB inhomogeneity correction algorithms to accurately model dose attenuation by high-density materials.

Calculation points placed within femurs were observed to have dose differences that varied case-by-case. The wide range of percentage dose differences and lack of systematic dose outputs were likely caused by the aforementioned deficiency of the PB inhomogeneity correction algorithm. Heterogeneity in the femur is especially prominent, with lower densities towards the center of the bone tissue. RadCalc®'s inhomogeneity correction algorithm was not able to effectively account for this, as it assumed uniform density within structure. Thus, depending on the density distribution of the femur bone, both overestimations and underestimations of dose were possible, as indicated by the results. Due to the inconsistency of these estimations, it is difficult to assign one single density to the femur contour which will correctly calibrate the dose calculation to any given calculation point within the femur.

In this study, the doses computed by the TomoHDA™ TPS were used as the reference standard for comparison. All the plans were approved for clinical delivery and had undergone verification by patient-specific DQA using ArcCHECK (Sun Nuclear Corp. Melbourne, FL, USA) and A1SL ion chamber (Standard Imaging, Middleton, WI, USA) measurements. As such, RadCalc®'s tomotherapy calculations could be validated if a good agreement between the doses

calculated by each program was observed. By including both lung and prostate treatment sites, we were able to further evaluate the performance of the program in both homogeneous and inhomogeneous targets.

One limitation of the study is related to the calibration of RadCalc®. Machine reference output was collected independent of the TPS and entered into RadCalc® for calibration, but percentage depth dose (PDD), off-axis ratio (OAR), and output factors settings were left unaltered. While it was suggested by RadCalc® that the default data are consistent across all tomotherapy units and available machine data were compared with RadCalc®'s data with little-observed percentage difference, results of the study could be affected to a small degree. Furthermore, several variables were introduced into the study when preparing the patient plans for dose calculation. No standard exists for assigning densities for ROIs, as with the placement position of a calculation point within an ROI. Although minor to the overall dose computation, this may have created variability between the examined cases.

The following recommendations for further development of the RadCalc software were proposed:

1. Improvement and calibration of inhomogeneity correction methods to perform better in low-density tissue and to function in high-density tissue without outputting zero doses
2. Exploration of methods to better represent the density distribution within an ROI through consideration of CT image data
3. Improvement of the PB dose calculation algorithm to implement corrections for its tendencies to overestimate doses.

CONCLUSIONS

RadCalc®'s recently released tomotherapy module is still in its initial development phase for beta testing. Comparisons between the reference TomoHDA™ TPS doses and RadCalc® doses show deviations to be the greatest among high or low density and inhomogeneous regions, as expected for pencil-beam dose calculation algorithms. PTV calculation points for prostate cases were relatively consistent despite a systematic overestimation, most likely attributed to the PB dose calculation algorithm. The time required to prepare and calculate a plan using RadCalc® was much longer than the time taken to perform patient specific QA using a phantom and an ion chamber, which provides a more comprehensive check to a treatment plan. As such, while RadCalc®'s tomotherapy module is capable of estimating point doses in homogeneous target with a small degree of overestimation, it is currently not a preferred method of performing QA for helical tomotherapy.

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Nil.

Conflicts of interest

There are no conflicts of interest.

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