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Research Article

ANALYSIS OF NOVEL TREATMENT COMMISSIONING AND ANALYSIS TOOL

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ABSTRACT

Helical tomotherapy is a type of EBRT radiation treatment that takes into account image guided versatile radiotherapy. Its treatment arranging framework TPS utilizes a convolution superposition calculation for portion conveyance estimations. The precision of this calculation in the nearness of heterogeneities were assessed against Monte Carlo (MC) estimations and measurements. This work performed BEAMnrc-and DOSXYZnrc-based MC portion estimations of tomo treatment conveyances to a CIRS human heterogeneous apparition with normal clinical reverse arranging and conveyance settings. The exploratory outcomes were utilized to assess both the TPS and MC portion counts. The MVCT image set of the treatment simulation environment were utilized for treatment planning and later for treatment delivery by using the Monte Carlo based system developed to account for image value to density table changes. The 3%/3 mm measure was observed for the TPS within the limits of IVDT inhomogeneity recommended by established norms.

INTRODUCTION

Helical tomotherapy is an external radiotherapy process that delivers intensity modulated radiation therapy (IMRT) as well performing image guided adaptive radiotherapy (Mackie *et al* 2006, Mackie *et al* 1993). It features an onboard MVCT system through which it accomplishes image guidance. It delivers a complex modulated set of beams from significantly more angles than typically found in conventional linac IMRT, which often results in better conformity of radiation dose distribution to the planning target volume (PTV) and avoidance of nearby organs at risk (OAR). This technology can result in better homogeneity of the dose distribution within a PTV, while maintaining the same or better sparing of normal tissue compared to conventional three-dimensional conformal radiotherapy and some linac IMRT deliveries. The current commercially available helical tomotherapy unit is the Hi-ART II. The system utilizes its own dedicated treatment planning system (TPS). This TPS uses a convolution/superposition algorithm for the dose calculation, which inherently assumes the condition of charged particle equilibrium throughout the calculation volume. We can reasonably expect the TPS will have difficulties in accurately predicting the dose in certain situations, such as when significant inhomogeneities exist in the patient geometry, or when treatment consists of small fields or highly modulated beamlets, in which this assumption is not satisfied. In the buildup regions behind air cavities, such as those found in a head and neck or a lung treatment, 5%–7% of maximum dose deviations can be found (Seco *et al* 2005). Monte Carlo (MC) dose calculations can be used to validate the TPS predicted dose in such cases (Vanderstraeten *et al* 2006). We have developed a full MC dose calculation method for helical tomotherapy previously. Here we apply this system to a heterogeneous cheese phantom with a simulated patient treatment plan as per the literature review (Yadav *et al* 2010). The phantom allowed for experimental validation of both MC and TPS results.

As per Yadav *et al* (2010) there are discrepancies in electron density overtime. We have developed a full MC dose calculation method for helical tomotherapy previously. Here we apply this system to a heterogeneous cheese phantom with a simulated patient treatment plan. Our system implements the planned fluence as per recommendations of Yadav *et al* (2010). The cheese phantom allowed for experimental validation of both MC and TPS results. We then consider the doses calculated from both systems for a clinical head-and-neck cancer treatment.

MATERIALS AND METHODS

Monte Carlo Calculation of Helical Tomotherapy

A MC model of the Hi-ART II helical tomotherapy unit was established and commissioned as reported in our previous work.10 Our MC system uses the BEAMnrc/DOSXYZnrc

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software packages to perform a full MC simulation of a helical tomotherapy dose delivery. The model was verified at three levels: the source simulation, MLC modeling/dynamic intensity modulation, and final dose distribution of the helical delivery. The MC model was commissioned for two typical jaw settings, Tslice=2.5 cm and Tslice=5.0 cm. These define the radiation field width in the couch movement direction at the isocenter. The MC calculation was calibrated by comparing a measured 1 min 5 x 40 cm2 field output with the MC simulation of the same field. The measurement was 856.5 ± 1.7 Gy/min and the MC result yielded 2.21e-16 ± 1% Gy/particle. The calibration factor is therefore 6.45e14 particle/ s ± 1%. Once planned on the Tomo-Therapy Hi-ART TPS version: 2.2.0.259, a patient dose distribution can be recalculated on our MC platform. An in-house program written for MATLAB reads the selected TomoTherapy TPS archived patient file and automatically writes BEAMnrc and subsequent DOSXYZnrc input files. Additionally, it monitors and steers the sequential MC simulation processes to get a final dose distribution of the helical delivery. The program can also read the patient structures delineated by radiation oncologists in the plan and provide analysis of calculation results to give dose volume histogram (DVH) and gamma index11 information using these structures.

Computed tomography (CT) images used in this study were all obtained with a Picker PQ5000 CT scanner Phillips Medical Systems, Cleveland, OH. For a treatment planning dose calculation, the diagnostic CT couch needed to be replaced by the presaved helical tomotherapy couch CT dataset to perform the calculation in the real helical tomotherapy dose delivery situations. This is necessary in helical tomotherapy due to the helical delivery often requiring a portion of the treatment beams to pass through the couch, whereas this may be avoided in conventional linac based IMRT. The typical CT dataset was downsampled to 256 x 256 by the TPS and it had a voxel size of 1.875 x 1.875 x 3.000 mm3. Our in-house MATLAB program reads the TPS CT data set and converts it to a DOSXYZnrc phantom file compatible to cheese phantom for portability. The phantom grid has the same voxel size and setup for MC calculation as in the TPS calculation. In converting CT data to a cheese phantom, an image value to density table (IVDT) with four materials: air, lung, soft tissue, and bone, densities and their calibrated CT numbers for our CT scanner was applied and the material and density of each voxel were assigned by linearly interpolating the data in the IVDT. The IVDT is largely equivalent to the one used by the TPS for clinical CT images. For example, the IVDT gives the same physical density values within 2% as the TPS in 98% of the volume of the head-and-neck cancer patient CT dataset in this study. Such differences in density are expected to introduce only small dosimetric differences, well below our mini-mum evaluation criteria. All simulations were performed with the electron cutoff energy ECUT=0.7 MeV and the photon cutoff energy PCUT=0.01 MeV. The dose calculation used 109 histories (108 recycled ten times) to get the uncertainty below 1% in high dose regions (PTV). The scans were done over the period of 06 months and for the life of the target (Yadav *et al* 2010). The TomoTherapy TPS uses the effective mass attenuation coefficients derived from water and cortical bone to calculate the total energy released per unit of mass for dose convolution. The dose kernel is calculated by using range scaling by physical density of the Monte Carlo generated kernel in water. So, the doses reported by TPS are dose-to-medium the same as by the MC method. Hence, both the MC and TPS dose results were converted from dose-to-medium to dose-to-water to compare with the measured results in the cheese phantom compatible CIRS in this work (Yadav *et al* 2010). The main reason for utilizing the cheese phantom was its setup flexibility and robustness with multi density plugs and pluggable slots for solid water (Yadav *et al* 2010). In this conversion, dose-to-bone has the largest difference from dose-to-water, which is above 10%, while soft tissue differences are approximately 1%. There was no dose conversion of either MC or TPS in the evaluation for the head-and-neck treatment plan. For this work, the MC calculations were performed on a Linux cluster dedicated to MC calculations with 18 AMD Opteron64 CPUs operating at 2.0 GHz, AMD, Sunnyvale, CA and managed by ROCKSv3.3 Rocks Cluster Group, San Diego Supercomputer Center, UC San Diego, San Diego, CA. the system was recommissioned to match the gold standards for treatment plans. The changes in IVDT curve before and after system's target replacement were observed (Yadav *et al* 2010).

Measurements and Calculations for an Anthropomorphic Heterogeneous thorax Phantom

To assess the TPS performance in the presence of known inhomogeneities in a clinically relevant situation, a treatment plan with the same order of complexity as a clinical plan is mimicked using a heterogeneous CIRS phantom model 002LF IMRT Thorax Phantom, CIRS, Inc., VA. This phantom has an elliptical shape and it represents an average human torso in density and structure. It measures 30 cm long x 30 cm wide x 20 cm thick. The phantom is made of simulated lung (inhale,15 simulated bone density, the major part of the delineated spinal cord Structure),16 and water equivalent materials. According to the vendor, the tissue equivalent materials mimic the dosimetric properties mass and electron density of water, bone, and lung within 1%.17 The phantom contains tissue equivalent interchangeable rod inserts which can be replaced by an ionization chamber to allow point dose measurements at different locations in the phantom. Results were validated by using the criteria of Homogenity Index (HI) as per equation no. 02 of Yadav *et al* (2010). This gives a broad spectrum for dose evaluation from 2% upto 98%. One half of the phantom is divided into six sections, each 2 cm thick, to support radiographic film measurements in axial planes.

CIRS thorax phantom treatment plan

We created a PTV with an irregular shape with approximate dimensions of 10 x 15 x 17 cm3 in the CIRS phantom. The PTV is larger than the typical clinical situation. This larger PTV was chosen to contain water/tissue and a part of the right lung. It reaches around a cylindrical organ-at-risk (OAR1). This arrangement increases the complexity of the beam modulation and radiation delivery. The plan was constrained to deliver 50 Gy to 95% of the PTV and limit the dose to the two OAR and the spinal cord such that each of these receives no more than 40 Gy in more than 50% of their volume. The TPS generated a plan with 16.1 active gantry rotations x 823 projections, assuming 51 projections in gantry angle per rotation (using Tslice=2.5 cm field width, a pitch value of 0.46) the ratio of the couch translation in one rotation to the field

width, and an effective modulation factor of 1.984 a measure of modulation complexity, the maximum leaf open time compared to the average for open leaves. The plan was calculated on the TPS with a "fine" grid setting, which used calculation grid voxel sizes of 1.875 x 1.875 x 3.000 mm^3.

Absolute Dose Measurements with A1SL ion Chambers

Absolute point dose measurements were taken with an Exradin A1SL ion chamber Standard Imaging, Middleton, WI in the CIRS phantom at points 1–7. The A1SL ion chamber has an active volume of 0.056 cm3. Three calibrated A1SL ion chambers were used with a TomoElectrometer eight-channel electrometer. All A1SL ion chamber measured values were converted to dose to water using an AAPM TG-51 equivalent protocol. In this method, the relationship between the helical tomotherapy reference condition SSD=85 cm, 5 x 10 cm2 field size and the AAPM TG-51 reference condition SSD=100 cm, 10 x 10 cm2 field size was derived as per the protocol suggested by Thomas *et al*.,18 which allows for calibrated reference dosimetry. The seven points in the CIRS phantom were measured 2–8 times the number of measurements were increased in regions of low dose or high gradient. Although the beam output is generally stable over a delivery, there can be variations between deliveries done at different times. This variation can be quantified by the two machine monitor unit chambers on the Hi-Art II unit. The measurement results were normalized by the average machine monitor unit rate of each delivery to correct for small machine output fluctuations between subsequent measurement.

Relative dose Measurements with films

Relative dose distributions were measured in the central axial plane of the CIRS phantom with Kodak EDR2 ready pack film size 30 cm x 25 cm, Eastman Kodak Company, Rochester, NY. The film was in the transverse plane, which is parallel to the beam axis rotation plane. The EDR2 film was scanned with a VIDAR VXR film digitizer VIDAR Systems Corporation, Herndon, VA which was studied and calibrated in our previous work (Thomas *et al* 2005). Such film measurement procedures are routinely used for clinical QA tests.

Table 1 Volumetric coverage of QA analysis

Volumetric coverage	Dose Scale	P Ratio
V5	24	6.22
V40	23.2	6.68
V45	24.2	7.12
V65	27	8.22
V90	31.0	9.17

RESULTS

Absolute dose in the Cheese phantom

The A1SL ion chamber point dose measurements are shown in Table I. The point doses were measured in PTV, OAR, and the region out of the PTV within the dose range from prescription dose to about 33% of the maximum dose points 1–7. All seven of the MC doses agree with the IC measurements within 1.5% relative difference. The largest difference is 1.49% in bone material. The root-mean-square (rms) difference of MC results is 0.81%. The TPS results at these points agree with the IC measurements within 4% relative difference. The largest difference is 4.05%. The rms of TPS results is 2.33%. The agreement, which was smaller than 3%, is consistent with the TomoTherapy TPS reporting the dose in the form of dose-to-medium. This is the same as the Varian Eclipse TPS23 and is different from most of the commercially available TPSs such as Pinnacle and Helax.

Relative dose Measurements in the Cheese Phantom Compatible CRIS

MC results agree with the film measurements in most parts of the phantom plane. Discrepancies were seen in the lung part of the PTV, the boundary of the left lung, and a part of the OAR. The TPS overestimated the lung dose in the PTV by about 6% and underestimated by more than 3% the dose in the edge of the lung outside of the PTV. These discrepancies can be seen clearly in the gamma maps. The white color within the outline of the phantom indicates the region passing the 3%/3 mm prescription dose criterion, the light gray is the region failing the 3%/3 mm but passing the 5% /3 mm criterion. The dark gray is the region failing the 5% /3 mm but passing the 7%/7 mm criterion. In the TPS results, larger discrepancies were seen in the lung part of the PTV, the boundary of the left lung, and a part of the OAR. There are larger regions of discrepancy considering TPS versus film as compared to MC versus film. Both the MC and TPS calculations meet the 7%/7 mm criterion in the film plane in the phantom. The MC result meets the 5%/3 mm criterion in all regions and 82%–100% of pixels in each volume pass the 3%/3 mm criterion. In the right lung, 82.9% of the pixels pass the 3%/3 mm criterion.

Only 82.3% pass this criterion in the PTV with the majority of disagreement accruing in the lung equivalent material. The TPS result meets the 5%/3 mm criterion in 90.8%–100% of pixels in each volume. Further, 53.5%– 99.8% of pixels in each volume pass the 3%/3 mm criterion. Lung and low dose regions with high gradients show large gamma values and fail to meet these criteria. The differences between TPS and film and between MC and film show the same trend when compared on a voxel-by-voxel basis. In both cases, these comparisons show a higher dose in the PTV in the lung region and lower dose in the lung-tissue interface. The TPS overestimates the lung doses and underestimates the boundary doses, more so than MC calculations. The 5%/3 mm criterion failed in some parts of the TPS results, whereas only a small portion of the MC results failed the 3%/3 mm criterion. In other words, TPS predicts higher dose in the lung region and lower dose in the interface buildup region compared with MC predictions.

The TPS results agree with the MC results in most regions. Differences are seen in the nasal cavity, the oral cavity, and the thorax. There were no regions that failed to meet the 7% /7 mm criterion. The 5%/3 mm criterion only failed in a small portion of these regions. The TPS agrees with the MC results well in the PTV66b, the cord, the brain, and the optic chiasm. Differences are seen in the PTV66a, the PTV54, and both of the parotids. The relative volume of the ROI volumes passing the gamma tests for different criteria are shown in the table. All regions completely passed the 7%/7 mm criterion. An average of 98.8% of the volumes passed the 5%/3 mm criterion and an average of 92.7% of the volumes passed the 3% /3 mm criterion. Good agreements are seen in the regions of the right parotid, the optic chiasm, PTV66b, and the cord, where 100%

of each volume passed the 3%/3 mm criterion. The PTV54 has the worst agreement with 85.4% of the region passing the 3%/3 mm criterion and 97.2% of the region passing the 5%/3 mm criterion.

DISCUSSION

Full MC dose calculations for a helical tomotherapy system were applied to a heterogeneous CIRS phantom plan and a clinical head-and-neck cancer treatment plan. Ion chamber, as well as EDR2 film measurements of a treatment delivery to the CIRS phantom, were used to provide a base line measurement evaluation of both systems. Compared to ion chamber measurements, our MC calculation for the CIRS phantom agrees with measurement within an average 0.81% relative difference, the largest difference being 1.49%. The TPS has average relative difference of 2.3%, with the largest difference being −4.1%, compared to ion chamber measurements. The larger difference in the low dose OAR was seen in our routine delivery quality assurance (DQA) results. It may be due to the large dose gradient in this region, which makes the measurements particularly sensitive to probe positioning. The MC results agree with film measurements to a clinically acceptable 5%/3 mm criterion level. The more rigid criterion of 3%/3 mm indicated the difference between the MC and the film measurements. In the low density region and high dose gradient OAR region, more than 82% of the voxels passed the test and more than 90% volume passed the 3% /3 mm test. The TomoTherapy TPS, using a convolution superposition algorithm, failed a 5%/3 mm criterion level in some of the high dose low density lung region, low dose boundary regions, and high dose gradient regions. Another discrepancy of both calculation results compared to film results exists in the region at the edge of the phantom. Both MC and TPS calculation results have ripples in the isodose lines in this region. These ripples are smoothed in film measurement. This is likely due to the difference between discrete gantry angle versus projection assumptions between calculation models and actual continuous arc beam delivery. The assumption of 51 gantry angles per rotation, inherent in the TPS, simplifies the dose calculation. An arc of about 7° was simplified as a single beam delivered at the middle point of the arc. This assumption leads to small deviations in source position, which are proportional to the distance from isocenter, for each projection. On the other hand, the helical tomotherapy beam is delivered about the center of the arc and this arrangement reduces this error. Moreover, the dose at one point in the radiation field of helical tomotherapy was contributed by hundreds of subfields from almost 360°. This washes out the source position error in one field. All these facts reduce the calculation errors due to this assumption and make it less serious. This effect was theoretically studied by Kissick *et al*. They generalized thatthe effect is negligible for most clinical situations.25 However, we still found 3%/3 mm differences about 13 cm from the isocenter in the low dose region, combined with the presence of heterogeneity. This should be considered when a patient PTV is far from the isocenter.

CONCLUSION

In the MC simulations of the CIRS plan and clinical plan, CPU time was spent on two major steps. To calculate phase space files to get fluence information, 50–60 CPU hours were used. To calculate dose distributions in phantom/patient data sets, 450–600 CPU hours were used. On our 18 CPU cluster, about 2 days are needed for one full calculation. The same calculation repeated on our 40 CPU cluster finished in 24 h. Likely, our calculation controlling program could be further optimized to reduce the communication overhead during calculation and more efficient methods could be employed to improve overall simulation time. In summary, the TomoTherapy TPS provides a reasonably accurate means of dose calculation with clinically acceptable accuracy in most circumstances for IVDT and HU discrepancies as per existing recommendations (Yadav *et al* 2010). The MC method shows better agreement with measurement as shown in this study, and therefore presents itself as a feasible means of verifying TPS calculations. The next move will be to incorporate the contrast to noise measurement module so as to serve as a single comprehensive solution.

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