Original Article

Monte Carlo Modeling of Dynamic Tumor Tracking on a Gimbaled Linear Accelerator

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Abstract

Purpose and Aim: The Vero4DRT (Brainlab AG) linear accelerator is capable of dynamic tumor tracking (DTT) by panning/tilting the radiation beam to follow respiratory-induced tumor motion in real time. In this study, the panning/tilting motion is modeled in Monte Carlo (MC) for quality assurance (QA) of four-dimensional (4D) dose distributions created within the treatment planning system (TPS). **Materials and Methods:** Step-and-shoot intensity-modulated radiation therapy plans were optimized for 10 previously treated liver patients. These plans were recalculated on multiple phases of a 4D computed tomography (4DCT) scan using MC while modeling panning/tilting. The dose distributions on each phase were accumulated to create a respiratory-weighted 4D dose distribution. Differences between the TPS and MC modeled doses were examined. **Results:** On average, 4D dose calculations in MC showed the maximum dose of an organ at risk (OAR) to be 10% greater than the TPS' three-dimensional dose calculation (collapsed cone [CC] convolution algorithm) predicted. MC's 4D dose calculations showed that 6 out of 24 OARs could exceed their specified dose limits, and calculated their maximum dose to be 4% higher on average (up to 13%) than the TPS' 4D dose calculations. Dose differences between MC and the TPS were greatest in the beam penumbra region. **Conclusion:** Modeling panning/tilting for DTT has been successfully modeled with MC and is a useful tool to QA respiratory-correlated 4D dose distributions. The dose differences between the TPS and MC calculations highlight the importance of using 4D MC to confirm the safety of OAR doses before DTT treatments.

Keywords: Dynamic tumor tracking, four-dimensional dose calculations, *Monte Carlo,* motion management

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Introduction

During the delivery of radiation therapy, intrafractional tumor motion poses a serious challenge to maximizing tumor coverage while sparing nearby organs at risk (OARs).[1] Respiration, for example, can cause significant organ and tumor motion in the thorax and abdomen, particularly near the diaphragm.[2] Managing respiratory-induced-tumor motion is especially crucial when delivering stereotactic ablative radiotherapy (SABR), a technique using highly conformal biologically effective doses with steep dose gradients. Respiratory-induced motion during SABR treatments can lead to target underdosage or OAR overdosage,[3] which may have clinical implications. Therefore, motion management techniques such as real-time dynamic tumor tracking (DTT), gating, or motion-encompassing target margins are essential when treating lesions near the diaphragm (e.g., liver) with SABR due to the degree of motion in this region.[2]

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The Vero4DRT (Brainlab AG, Germany) linear accelerator (linac) can perform respiratory-correlated DTT via its gimbal-mounted linac head that can pan and tilt the radiation beam up to ± 2.4° in two orthogonal directions, allowing tumor tracking anywhere within a ± 4.2 cm region in the isocenter plane.[4] A four-dimensional (4D) respiration correlation model is created between the motion of internal fiducial markers and external infrared (IR) markers on the abdomen. By monitoring the motion of the external markers during treatment, the location of the tumor can be predicted and tracked.[5-7] The Vero4DRT has an O-ring gantry that

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rotates the linac head around a patient's inferior-superior axis and posterior-anterior axis. It also has two integrated and orthogonal sets of kV X-ray tubes and flat panel detectors for on-board imaging.[5-7]

Currently, DTT treatment plans for the Vero4DRT are created, optimized, and evaluated on a single respiratory phase computed tomography (CT) image only, as there is no commercial treatment planning system (TPS) that can model the time-dependent panning and tilting motion of the treatment beam. Therefore, the changing source-to-target and source-to-surface distances during DTT are unaccounted for. In addition, the relative distance between OARs and the target as they move and deform during the respiratory cycle is not considered in the three-dimensional (3D) TPS calculations. This may leave an OAR vulnerable to exceeding its planning dose constraint while being treated during other breathing phases. Depuydt *et al*.[8] addressed this planning limitation by recalculating a 3D treatment plan that was optimized on one breathing phase onto other respiratory phases by approximating the beam's panning/tilting geometry as a simple translation rather than a proper rotation. Prasetio *et al*.[9] modeled the proper rotation of the beam when it pans and tilts by rotating the CT image for each breathing phase using an in-house developed software, and then recalculating the dose distribution.

Previously,[10] a novel method was developed for accurately modeling panning and tilting while recalculating a plan on multiple breathing phases of a patient's breathing cycle and accumulating these dose distributions into a single 4D dose distribution within the TPS. This is accomplished by monitoring the displacement of the target from the original planning CT to each phase of the 4DCT (based on identifying the location of the fiducial markers implanted around the target). The initial gantry and ring angles of each beam from the original plan are transformed into a new set of gantry, ring, and collimator angles, and the patient is shifted along the beam path, so as to re-create that beam's direction and length in the body during each phase of the respiratory cycle. Each beam in the original plan undergoes this transformation, and the new beams are saved to a new treatment plan on the appropriate respiratory phase. The 3D dose distribution can then be recalculated on each phase image. Using deformable image registration (DIR), the dose distributions on each phase of the 4DCT can be deformed to the original planning CT image and accumulated to create a single 4D dose distribution. This beam and patient coordinate system transformation method is described in detail in previous work.[10]

A 4D dose distribution is necessary for making well-informed clinical decisions about DTT treatments because they provide more accurate dosimetry than a simple 3D dose calculation on a single breathing phase. Implementing a secondary dose calculation safety verification check on the 4D calculations can be challenging. Monte Carlo (MC) modeling is often used for patient-specific TPS quality assurance (QA), as it is considered the "gold-standard" for dose calculations.[11,12] Currently, at the author's center, MC is used for secondary verification of all 3D sIMRT dose distributions without modeling the panning and tilting of the treatment beam. To use MC to verify the 4D dose calculations developed for DTT treatments, it is necessary to model panning and tilting in MC and to then accumulate the 4D dose calculated on multiple breathing phases onto a single reference phase. One method of modeling the panning and tilting in MC has been previously described by Ishihara *et al*.[13] In their approach, the MC phase space data are rotated after transport through the linac head to simulate the panning/tilting of the Vero4DRT's beam during the dose calculation. The dose to OARs calculated on a single reference respiratory phase is compared with the dose to that OAR on each individual phase of the 4DCT. Their approach has two limitations: (1) It does not accumulate the 4DCT doses to one reference phase to create a single 4D dose distribution, which limits the potential of their study for clinical decision making; (2) their technique for modeling panning and tilting cannot be implemented in the clinical TPS. Therefore, it can only be used during the final step of plan QA, and cannot contribute feedback to the user during the plan optimization stage.

In this work, the modeling of the beam panning and tilting motions in the MC system is investigated and a method for accumulating dose from multiple phases into a 4D dose distribution is described. This MC model is applied to 10 patient plans optimized during previous work[10] to demonstrate the necessity and suitability of MC for patient-specific QA for DTT plans. The MC QA methods described here address both of the limitations described in the Ishihara *et al.* implementation: (1) Deformable dose mapping is used to accumulate the MC doses from multiple phases on a reference image to achieve a single 4D dose distribution that provides more meaningful clinical information; (2) the methods for modeling panning and tilting and deformable dose mapping are the same in MC as those implemented in the TPS. Therefore, the 4D dose calculations provided by the TPS can be directly compared with MC using the same modeling methodology. The workflow followed in this study is shown in Figure 1. This work demonstrates for the first time that the BEAMnrc code can support and successfully simulate the panning and tilting of a radiation beam. This opens the door to future MC studies that involve modeling the panning/tilting of the beam. This work also explores differences between the TPS and MC with hypotheses as to why discrepancies may exist; to the authors' knowledge, this is the first time MC and TPS 4D dose accumulations have been compared.

Materials and Methods

Modeling a four-dimensional panning/tilting radiation beam in the treatment planning system

This retrospective study used CT data from 10 previously treated liver SABR patients. All 10 patients had three or more gold fiducial markers implanted near the tumor before a

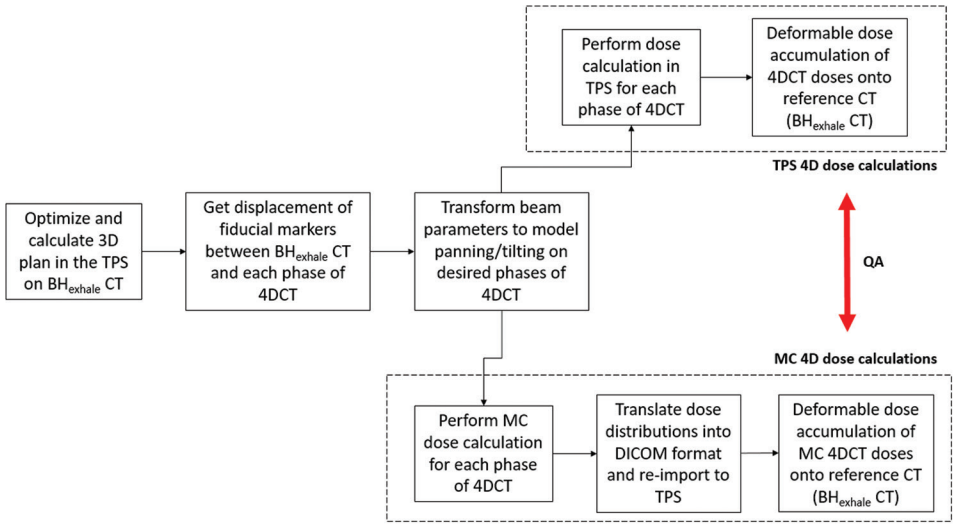


Figure 1: Workflow for 4D MC QA. The workflow for QA of DTT plans that are optimized on a single phase in the TPS. The transformation used to alter the original plan's beam parameters to model panning and tilting is the same for both 4D dose calculations in the TPS and 4D dose calculations with MC. The dashed boxes around certain parts of the workflow indicate which steps are for 4D dose calculations with the TPS or MC specifically. The 4D MC dose calculations can be used for QA of the 4D dose calculations produced in the TPS, or can also be used for QA of the original plan optimized on the Bh_{exhale} CT in the TPS. QA: Quality assurance, DTT: Dynamic tumor tracking, TPS: Treatment planning system, MC: Monte Carlo, CC: Collapsed cone

planning CT simulation scan. A 7-beam, 6 MV step-and-shoot intensity-modulated radiation therapy (sIMRT) treatment plan was created and optimized on each patient's breath-hold CT image taken at exhale (BH_{exhale}) in the RayStation TPS (RaySearch Laboratories, Sweden), which uses the collapsed cone (CC) dose calculation algorithm (v5.1). Using a script developed in-house[10] this treatment plan is transferred to all 10 phases of the 4DCT and each beam's gantry, ring, and collimator angles are transformed for every breathing phase to model panning and tilting. In addition, the patient position relative to the source is also shifted to account for the changing beam path length when the beam pans/tilts. A radiation oncologist contoured OARs of interest [1–3 OARs per patient, Table 1] on all 10 4DCT phases and the BH_{exhale} CT used for planning. The results of this work have been previously published.[10]

Monte Carlo model

An MC model of the Vero4DRT has been built in-house and validated for dose distribution simulations without accounting for DTT beam panning and tilting motions.[14,15] The EGSnrc/BEAMnrc code was used to simulate the 6MV photon beams.[16,17] To obtain a dose calculation uncertainty of <2%, 1.5×10^9 electrons were incident on the Vero4DRT target, and the resulting bremsstrahlung photons and scattered electrons were transported through the linac head, including the primary collimator, flattening filter, ionization chambers, and stationary secondary collimators. All relevant information about the components of the linac head were provided by the developer.[18] A phase space plane was created below the static secondary collimator and above the dynamic multi-leaf collimators (MLC's), which includes the photons and electrons transported through the linac head (i.e., the nonpatient-specific components of the Vero4DRT). Particles were then directed through the MLCs and the patient's CT image using a shared library format of BEAMnrc code combined with "source20" of DOSXYZnrc code.[14,15] The photon cutoff energy used was 0.01 MeV and the electron cutoff energy was 0.521 MeV. All simulations were conducted using the Condor High-Throughput Computing software[19] on a cluster with eight servers with the Red Hat Enterprise Linux operating system (v. 6.4). These servers have two processors (Intel Xeon CPU E5-2650 0 @ 2.00 GHz) with 8 cores per CPU, and two threads for each core, providing 256 nodes for calculations.

Monte Carlo with panning and tilting

Panning and tilting geometry was modeled in MC using a series of equations that transform each beam to correspond with the panning and tilting angle that correlates with the respiratory phase image dose is being calculated on.[10] These equations alter each beam's gantry, ring, and collimator angles and shift the patient based on knowledge of the displacement of the fiducial markers between the BH_{exhale} CT and the phases of the 4DCT dose is being calculated on. The transformation re-creates the beam's panning/tilting path through the body during DTT with an accurate source-to-surface and source-to-target distance.

Four-dimensional Monte Carlo dose distributions

The original treatment plan, optimized for the BH_{exhale} CT using the TPS' CC algorithm, was recalculated on the BH_{exhale} CT in MC. Similarly, the modified plans that model panning and tilting for each of the 10 phases of the 4DCT were calculated with MC. All dose calculations, both with MC and CC, use a 2 mm × 2 mm × 2 mm dose grid. The 11 total dose distributions per patient were converted from MC text format into a standard DICOM format and re-imported to the TPS where they could

Table 1: The maximum dose (D_{max}) for all organ at risk of interest in this study among 10 patients

Maximum dose of OARs in patient study using different calculation methods						
Patient number	**OAR**	**Dose limit**	**D_{max} on BH_{exhale} (CC)**	**D_{max} on BH_{exhale} (MC)**	**D_{max} from 2-phase 4D MC**	**D_{max} from 10-phase 4D MC**
1	Duodenum	3200	3039	3132	3802	3740
	Large bowel	3800	3731	4288	4558	4601
	Chest	5500	5462	5519	5843	5867
2	Stomach	2220	1186	649	631	611
	Large bowel	2820	1175	1281	1297	1372
	Heart	3000	2592	2272	2127	2003
3	Duodenum	2220	92	144	138	257
	Heart	3000	2115	2161	1667	1904
4	Duodenum	2220	2201	2303	4428	3950
	Major vessels	4400	4336	4615	4410	4353
	Large bowel	2820	2651	2797	2645	2619
5	Large bowel	3800	1895	2147	2130	2096
	Major vessels	5150	3360	3408	3049	3022
	Heart	3800	3608	3671	3179	3015
6	Major vessels	4400	1592	1659	1687	1425
	Duodenum	2220	1557	1589	1944	1896
7	Major vessels	4400	3033	3267	4043	4042
	Duodenum	2220	831	908	1424	1388
8	Stomach	2220	877	998	1171	942
	Large bowel	2820	1684	1876	1066	665
	Duodenum	2220	761	821	655	635
9	Chest wall	4400	2866	2854	2508	2561
10	Esophagus	2700	2605	2597	2105	2124
	Spinal canal	2030	1982	2070	2219	2111

The dose limits follow published dose constraints, and all dose values tabulated are reported to 0.03cc volume. The shaded rows are OARs that exceeded their dose limit using an MC calculation. All dose calculations used a 2 mm×2 mm×2 mm dose grid. OAR: Organ at risk, CC: Collapsed cone, MC: Monte Carlo

be conveniently viewed and compared on the corresponding CT image. The ten 4DCT respiratory phase dose distributions were deformed to the BH_{exhale} CT using the ANACONDA DIR algorithm[20] available in the RayStation TPS. Using the BH_{exhale} CT as the reference phase, two 4D dose distributions were created from MC calculations:

1. 10-phase 4D MC: All ten phases from the 4DCT were summed together
2. 2-phase 4D MC: A simplified version where the 0% phase and 50% phase (inhale and exhale, respectively) dose distributions were summed together with patient-specific respiratory phase weightings.

While summing plans from all 10 breathing phases is the most accurate, it requires significantly more time because each phase needs to have the OARs of interest contoured to improve the accuracy of the DIR workflow and there is additional calculation time required on the MC cluster (five times more plans need to be calculated). Therefore, the results from a simplified 2-phase 4D dose distribution are compared to the complex 10-phase 4D dose distribution to determine if using only the two extreme breathing phases provides adequate results in a more efficient time frame.

The respiratory phase dose distribution weightings were determined from the patients' breathing traces that were acquired during their 4DCT scans. The breathing trace amplitudes were divided into an upper "inhaling" half and lower "exhaling" half. The time spent in each half was used to determine the weighting between the inhale and exhale phases. An example of a patient's breathing trace divided based on inhaling/exhaling is shown in Figure 2. In addition to comparing MC and the TPS' CC 4D dose distributions, the dose distributions calculated on the BH_{exhale} CT by CC and MC were compared to investigate discrepancies between the two dose calculation algorithms.

Water-density patient computed tomography

To further explore and characterize the source of discrepancies between MC and CC algorithms, a 7-field sIMRT treatment plan was optimized on a single patient CT image using CC. The body was converted to water density before optimization to eliminate any heterogeneities. Several simplified spherical and cylindrical region of interest (ROIs) were contoured in the water body and systematically placed around the target to act as "OARs" in the high-dose region. The plan's dose distribution was recalculated in MC on the same CT image with the body set to water density. The maximum doses in the ROIs calculated by CC and MC were compared to investigate differences between the two dose calculation algorithms that are not

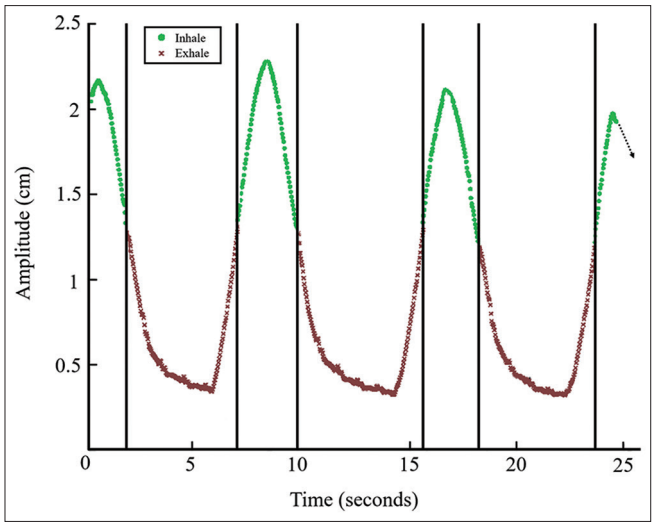


Figure 2: A patient's breathing trace. A patient's breathing trace is divided into cycles between points of maximum inhalation. Half the maximum amplitude in each cycle divides between inhalation (green circle) and exhalation (brown "x"). The time spent inhaling ($t_{in} = t_{in,1} + t_{in,2} + t_{in,3} + t_{in,4}$+…) and exhaling ($t_{ex} = t_{ex,1} + t_{ex,2} + t_{ex,3}$+…) from all cycles is used to determine the patient-specific inhale/exhale phase weighting

due to heterogeneous tissues, such as dosimetric differences occurring in the beam penumbra region.

Results

Four-dimensional Monte Carlo dose distributions

In this work, modeling of the Vero4DRT's panning and tilting beam with the appropriate coordinate transformation of the 3D treatment plan was successfully implemented into MC. The maximum dose to 24 OARs was examined among the 10 patients using 3D MC dose calculations (on the BH_{exhale} CT) and 4D MC dose calculations (2 phases only and all 10 phases). These distributions were compared with distributions calculated by the TPS' dose calculation algorithm, CC. All OARs were below their dose limit when optimized on the BH_{exhale} CT by design. When the plan was recalculated with MC on the BH_{exhale} CT, the calculated maximum dose to OARs was 5% greater on average (Ranging from -45% to 57%) than the maximum dose calculated using CC. Five OARs exceeded their planning dose limits when recalculated with MC on the BH_{exhale} CT, and the maximum dose was 6% higher on average (ranging from 1% to 15%) than what was calculated on the BH_{exhale} CT in the TPS.

4D dose calculations with 2 and 10 phases using MC showed the dose increased by a mean of 9% (Ranging from -47% to 101%) and 10% (Ranging from -61% to 179%), respectively, relative to the CC BH_{exhale} CT dose alone. Six OARs exceeded their dose limit using a 2-phase dose calculation with MC; for these six OARs the mean maximum dose was 28% greater than the CC single-phase dose calculation. Five OARs exceeded their dose limit using a 10-phase dose calculation with MC with an average maximum dose increase of 28% relative to the CC single-phase dose calculation. The maximum dose to all OARs from every calculation method mentioned is recorded in Table 1.

Figure 3 shows examples of using 4D MC dose distributions that model panning and tilting to confirm the maximum dose to OARs. A treatment plan was optimized for each patient on the BH_{exhale} CT in the TPS, and all OARs were below their dose limits, as depicted by the blue, diagonal bars in Figure 3. Using the 4D dose calculation method that models panning and tilting, the dose distribution is recalculated in the TPS using two or ten breathing phases (shown by the orange, horizontal bars, and the green, hash-mark bars, respectively). Now, the maximum dose for these six OARs is exceeding their respective dose limits, as indicated by the black horizontal lines in Figure 3. These plans are then recalculated in MC on the BH_{exhale} CT (black diagonal bars), and by creating a 4D dose distribution using 2 and 10 breathing phases (purple, vertical bars, and red, checkered bars, respectively) that models the correct panning/tilting of the beam. The maximum dose to the 6 OARs from these MC calculations is shown in Figure 3 beside the original dosimetric results from the TPS. The maximum dose to the 6 OARs that exceed their dose limit with a simplified 2-phase 4D dose calculation was 4% greater on average using MC than the TPS (Ranging from -1% to 13%). The 5 OARs that exceeded their limit using a 10-phase 4D dose calculation were 3% greater on average using MC than the TPS (Ranging from -1% to 7%). The 4D CC data were previously reported[10] but is included in Figure 3 for comparison with the 4D MC dose calculations.

Water-density patient computed tomography

The water-body CT image test eliminates any potential differences between the two dose calculation methods (MC vs. CC) caused by heterogeneities in the body. The results from this test are shown in Figure 4. The ROIs positioned lateral to the target (sphere 1, sphere 3, sphere 4, and sphere 5) had <3% difference in maximum dose between CC and MC. ROIs that were superior and inferior to the target (sphere 2, cylinder 1, and cylinder 2) had a difference in maximum dose of 6.6%-25.7% between the two algorithms. It was noted that ROIs in this region are primarily receiving doses from the beam penumbras.

To further investigate dose in the penumbra region, dose profiles were created from a 10 × 10 cm^2 field irradiating a square water phantom, calculated by CC and MC, at depths of 5 cm, 10 cm, 15 cm, and 20 cm. These dose profiles and their penumbra regions are shown in Figure 5. Table 2 summarizes the percent dose difference between the CC and MC profiles in the penumbra regions at different depths. From the central axis to 80% of the maximum profile dose, there is good agreement between MC and CC. From 50% to 80% of the maximum profile dose, CC doses tend to be higher than MC, and at the edges of the profile, between 20% and 50% of the maximum dose, MC is greater. This cross-over can be visually observed from the zoomed-in inserts in the plots in Figure 5.

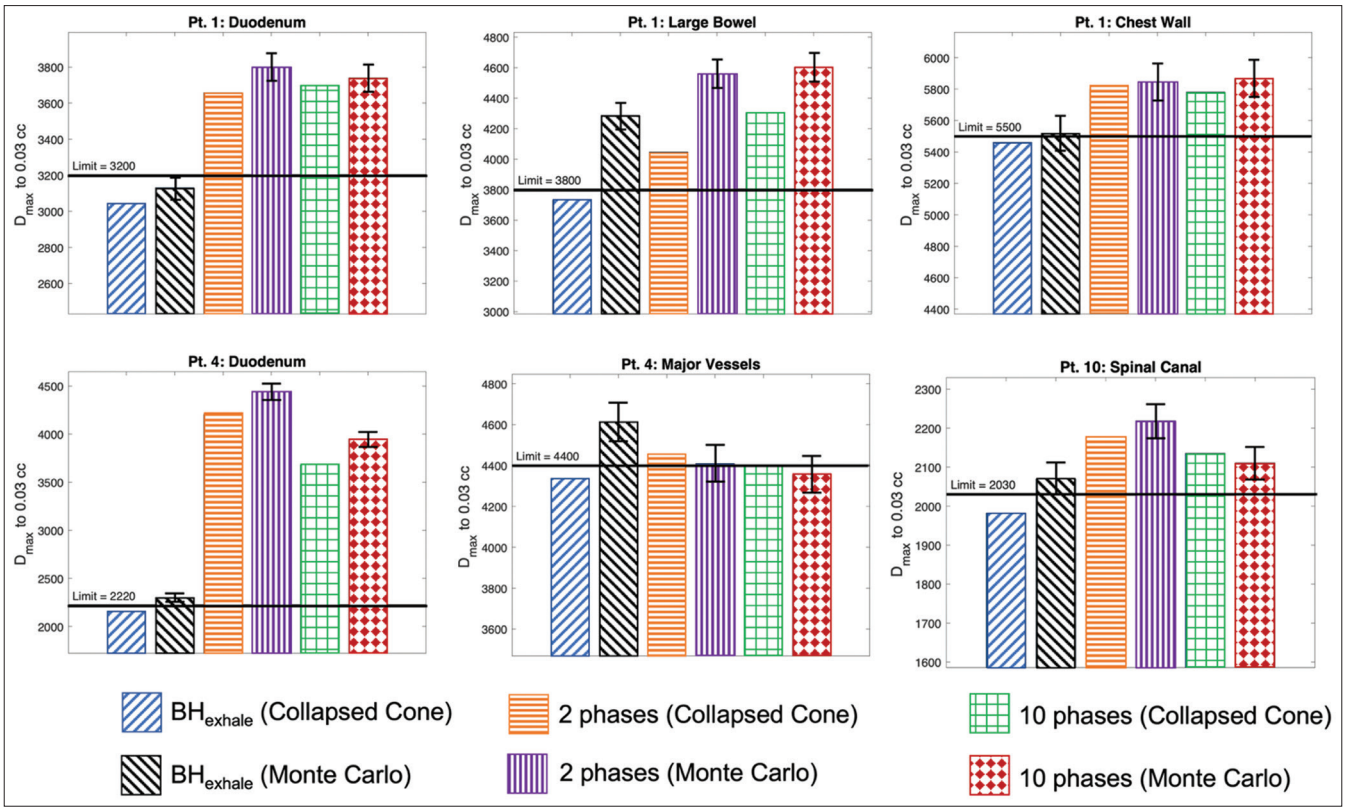


Figure 3: Maximum dose to OARs that exceed their dose limit using a 4D calculation method. This plot shows the maximum dose to 0.03cc of the six organs that met their dose limit when the plan was optimized on the BH_{exhale} image in the TPS, but exceeded their dose limit when the plan was recalculated using a 2-phase or 10-phase 4D dose distribution. The dose distributions were recalculated using MC, and the maximum dose to 0.03cc is reported for the 3D dose calculation on the BH_{exhale} image, and for a 2-phase and 10-phase 4D dose calculation. Error bars represent the 2% statistical uncertainty in the MC dose calculations. The horizontal black lines indicate the dose limit for that OAR. The exhale/inhale phase weightings for these three patients were 72%/28% (patient 1), 66%/34% (patient 4), and 65%/35% (patient 10). OARs: Organs at risk, CT: Computed tomography

Table 2: The mean, standard deviation, and range of the difference in dose between the CC and MC dose profiles at depths of 5 cm, 10 cm, 15 cm, and 20 cm [Figure 5]

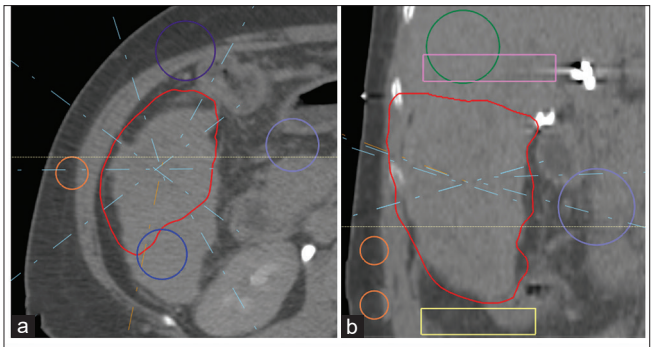
Depth (cm)	Dose differences between CC and MC profiles at different depths								
	80%–100% of maximum profile dose			50%–80% of maximum profile dose			20%–50% of maximum profile dose		
	Mean (%)	SD (%)	Range (%)	Mean (%)	SD (%)	Range (%)	Mean (%)	SD (%)	Range (%)
5	−0.3	0.5	−1.1–3.1	2.9	1.6	−0.9–5.6	−10.0	6.1	−21.7–0.7
10	−0.2	0.5	−1.2–3.8	1.4	1.9	−2.6–4.1	−9.4	3.8	−17.0–−2.2
15	−0.02	0.6	−1.2–3.2	1.5	1.0	−1.1–3.5	−10.7	7.4	−21.7–−0.3
20	0.06	0.6	−1.3–2.5	1.8	1.0	−1.0–3.6	−10.7	6.2	−22.0–0.1

The profiles are divided into three different regions: 20%–50% of the maximum profile dose, 50%–80% of the maximum profile dose, and 80%–100% of the maximum profile dose. The relative dose difference is calculated as (CC–MC)/CC×100. CC: Collapsed cone, MC: Monte Carlo, SD: Standard deviation

Discussion

A novel method for simulating a 4D dose distribution in MC that models the dynamic panning and tilting of the Vero4DRT treatment beam during DTT has been described. This method can be used for patient-specific QA of panning/tilting 4D dose distributions created in the TPS. The general methodology for modeling panning/tilting and accumulating the dose from multiple phases to create a 4D dose distribution is the same for both the TPS and MC. This allows for direct comparison of the results from both algorithms and thus can be used as a QA tool for 4D DTT dose verification. Although other groups have modeled panning/tilting in MC previously,[13] the ability to directly compare results with the TPS was not possible.

It is important to note that differences in the MC and CC dose calculation engines can lead to different results when reporting OAR dosimetry. These approximations can lead to differences between CC and MC's calculated dose, for example, in regions of heterogeneity, and in low-density regions with lateral charged particle disequilibrium.[21-23] When the original plan



ROI name	Collapsed cone max. dose (cGy)	Monte Carlo max. dose (cGy)	Dose difference: (CC-MC)/CC (%)
Sphere 1	5409	5264	2.7
Sphere 2	4166	3890	6.6
Sphere 3	2938	2920	0.6
Sphere 4	3732	3793	-1.6
Sphere 5	2515	2505	0.4
Cylinder 1	1346	1692	-25.7
Cylinder 2	3726	3342	10.3

Figure 4: Comparing maximum dose from CC and MC to OARs on water-density CT image. An (a) axial and (b) coronal cross section of the patient CT image set to water density (original CT numbers shown). The red contour is the target, and the colored spheres and cylinders are ROIs, labeled in the left column of the table. The dashed lines represent the center of the beams. The axial cross section in a) shows the ROIs lateral to the target (spheres 1, 3, 4, and 5) and the coronal cross-section in b) shows the superior/inferior ROIs (sphere 2, cylinder 1, and cylinder 2). The table below the images shows the maximum dose calculated for each ROI using CC and MC, as well as the percent difference of the two algorithms. OARs: Organs at risk, CC: Collapsed cone, MC: Monte Carlo, CT: Computed tomography, ROIs: Region of interest

on the BH_{exhale} CT was recalculated in MC, the maximum dose to some OARs were significantly different than what the TPS' CC algorithm calculated on the same image. In some cases, the maximum dose exceeded the OARs' dose limit, highlighting the necessity for re-calculating a plan's dose distribution in MC for QA purposes. For several OARs examined, the difference in maximum dose was even greater when comparing an MC 4D dose distribution (2 or 10 phases) to a CC 3D dose distribution. The changing source-to-surface and source-to-target distances, as well as OARs moving relative to the beam, can lead to different dosimetric outcomes during other breathing phases. Therefore, it is important to calculate a 4D dose distribution to make clinical decisions about the safety of a treatment plan before treatment.

The water phantom test results [Figure 4] showed a large difference in maximum dose between the two algorithms for the ROIs that were positioned superior and inferior to the target. Sphere 2, cylinder 1, and cylinder 2 were the only ROIs with a dose difference >3%, and as large as 25.7%. Since these ROIs were mostly superior and inferior to the incoming and exiting beams, they only receive dose from the beam penumbras and little to no contribution from the centers of the beams where the two algorithms have a better agreement.[21] Many OARs in the patient study that had a large maximum dose difference were also positioned superior or inferior to the target (for example, patient 1's large bowel).

Previous studies have confirmed that CC varies from MC in the beam penumbra region where there is lateral charged

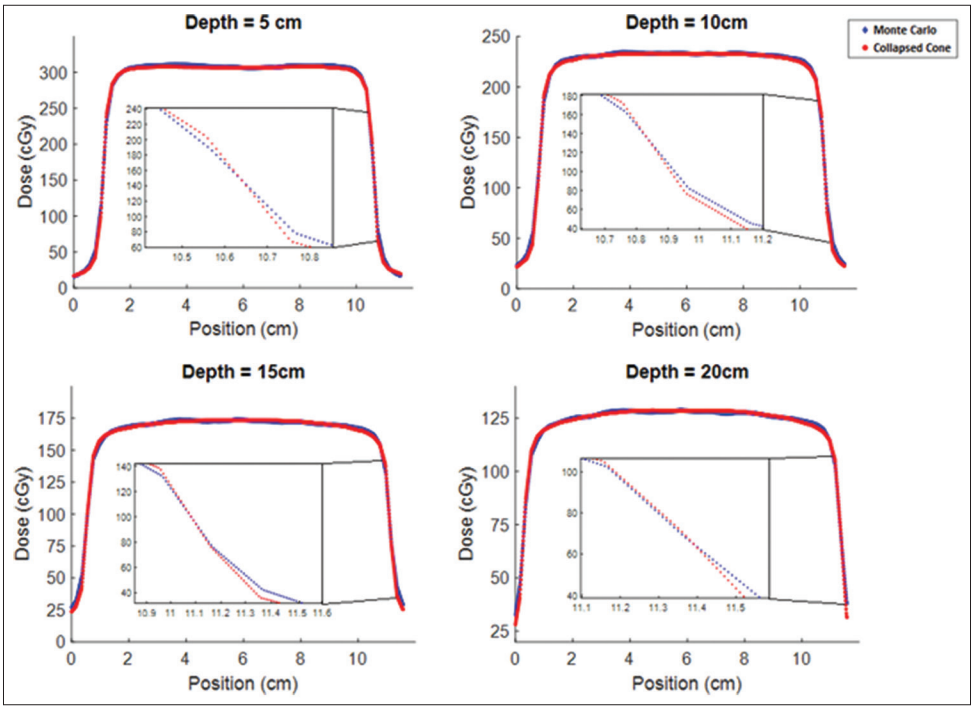


Figure 5: Dose profiles in water calculated with MC and CC at different depths. Dose profiles of a 6MV, 10 × 10 field at depths of 5 cm, 10 cm, 15 cm, and 20 cm in a water phantom were calculated with MC (blue diamonds) and CC (red circles). The insert under each profile shows a zoomed-in view of the portion of the penumbra between 20% and 80% of the maximum dose in the profile. MC data points have 2% uncertainty (not shown, too small). CC: Collapsed cone, MC: Monte Carlo

particle disequilibrium.[21,23] This explains why the largest dose difference between the two algorithms was seen for OARs positioned superior/inferior to the target, as they primarily receive a dose from the penumbra. In heterogeneous areas, the dose algorithms also perform differently.[21,23] However, most maximum dose points for OARs examined in this study are near the liver in soft tissue. Tissue heterogeneity is therefore not considered to likely be the greatest source of dose differences seen in the patient study.

The target of the radiation treatment will always be in the central portion of the beam during tracking since the beam is following the position of the tumor. Therefore, dose differences between CC and MC will be less prevalent for the target, and successful PTV coverage in the TPS will likely be confirmed by MC. However, OARs are generally receiving doses from the penumbra region of the beam since they are in the peripheries of the target. This study, as well as previous literature,[21-23] has shown that MC is more accurate at calculating dose in the beam penumbra than CC. Therefore, to calculate the accurate maximum dose to OARs in this area, it is essential to use MC as a QA tool. This further highlights the importance of confirming the safety of OARs during DTT treatments.

A major factor that contributes to dose differences between 3D and 4D dose calculations is changes in a patient's anatomy due to respiration.[1,2] For example, when a patient inhales, their liver moves mostly inferiorly towards other organs in the abdomen.[2] The dose to these organs will likely increase during inhalation because of the closer proximity to the target in the liver, and thus also to the beam during DTT treatments. Therefore, the dose in a 4D dose calculation would likely be greater because the inhale phase is included in the dose calculation, whereas the 3D dose calculations only model anatomy during exhalation (on the BH_{exhale} CT). Another example of changing anatomy influencing the calculated dose is random daily changes in anatomy. For instance, patient 4's duodenum had a maximum dose increase of 101% with the 2-phase 4D dose calculation compared to the dose calculated using CC on the BH_{exhale} CT. On closer inspection of the BH_{exhale} CT and the 0% and 50% phases of the 4DCT, a gas bubble can be seen entering the duodenum when the 4DCT is captured, that is not present when the BH_{exhale} CT image was taken approximately 7 minutes prior. This resulted in the duodenum moving closer to the high-dose region. These changes in anatomy are difficult to avoid, but in theory can occur for any patient during their treatment. In the results section, the average maximum dose increase between a 4D dose calculation and the original dose calculation using CC on the BH_{exhale} CT was reported for OARs that exceeded their dose limits using a 4D dose calculation. When the duodenum of patient 4 is removed from that analysis, the difference in the average maximum dose decreases from 28% to 14% (2-phases) and 23% to 12% (10-phases).

If one were to implement this MC workflow for QA of DTT plans for clinical purposes, there are some measures that can be taken to reduce time and resources. First, 2-phase 4D dose distributions produce similar results to a 10-phase 4D dose distribution and require significantly less time for contouring OARs of interest as well as MC computation time. This was also previously noted for 4D dose distributions created using CC.[10] Therefore, 2-phase 4D dose calculations would be sufficient and recommended. Furthermore, because MC is shown to have better accuracy than CC in the beam penumbra regions, one may choose to bypass creating 4D dose calculations in the TPS altogether. The new QA workflow would consist of only producing 4D dose distributions with MC to compare with the original 3D plan on the BH_{exhale} CT in the TPS to check the safety of OARs.

Figure 5 and Table 2 show the extent of dose differences in the beam penumbra region. While there is good agreement at the center of the beam, towards the far edges of the beam penumbra, there is an increased difference in the maximum dose. In the 20%–50% penumbra region of the beam, at depths of 20 cm, CC underestimates the dose by more than 10% on average, and at all depths, an individual point dose can be underestimated by up to 17%–22% in the TPS. These results in the 20%–50% region agree with previous studies that found CC underestimates dose outside the field.[21] In the water-density patient CT test [Figure 4], this trend is also demonstrated by the large difference in maximum dose for cylinder 1, which is receiving <50% of the prescription dose (4500 cGy). Dose differences in the beam penumbra may explain the significant dose differences found in the patient study [Table 1], as well as why the CC maximum dose was sometimes larger than MC's maximum dose, and vice versa.

One of the main limitations of using this method for QA of DTT plans is the assumption that a patient's breathing motion will be the same during treatment as it was when their 4DCT data was captured.[1] A possible solution to address this limitation would be to use the log files from the Vero4DRT after treatment to determine when the linac head was panning and tilting during treatment, and use this information to perform daily 4D dose calculations after each fraction to confirm if the dose delivered is consistent with what was expected. This may lead to applying offline adaptive re-planning if necessary. Another limitation is that, although this method is an effective QA tool for DTT plans, it offers no solution on how to proceed if the MC QA indicates an OAR is exceeding its limit and the plan needs to be reoptimized. To address this, it will be necessary to develop 4D treatment planning strategies that can be used during plan creation and optimization that takes into account information from multiple breathing phases.

Current and future work involves developing these 4D treatment planning strategies that incorporate multiple phases from the 4DCT, while modeling panning and tilting, into the treatment creation and optimization process, and not only during the plan evaluation phase. This will account for breathing motion during plan creation to deliver a more

optimal and safer plan for the patient. Another area of future investigation will be MC modeling of VMAT treatment plans incorporating panning and tilting using the methods discussed in this paper.

Conclusion

Correctly modeling the beam's panning and tilting geometry and accumulating dose from multiple breathing phases creates a 4D dose distribution that provides accurate dosimetric information for DTT treatments. MC modeling of this dynamic beam motion has been successfully implemented for the first time for 4D TPS dose calculation QA. The 4D dose distributions can be generated for a combination of all 10 breathing phases or for a simplified 2 breathing phase 4D dose distribution, depending on the resources available for contouring and MC calculation capacity. MC 4D dose calculations produce similar results to the TPS, with the biggest variation in maximum dose to OARs for organs that are superior/inferior to the target. These variations are expected based on the known differences between the two algorithms in the beam penumbra region, further highlighting the need for 4D MC QA calculations to ensure the safety of DTT treatments.

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Conflicts of interest

There are no conflicts of interest.

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