Development of a deformable phantom for experimental verification of 4D Monte Carlo simulations in a deforming anatomy

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Purpose: To verify the accuracy of 4D Monte Carlo (MC) simulations, using the 4DdefDOSXYZnrc user code, in a deforming anatomy. We developed a tissue-equivalent and reproducible deformable lung phantom and evaluated 4D simulations of delivered dose to the phantom by comparing calculations against measurements.

Methods: A deformable phantom consisting of flexible foam, emulating lung tissue, inside a Lucite external body was constructed. A removable plug, containing an elastic tumor that can hold film and other dosimeters, was inserted in the phantom. Point dose and position measurements were performed inside and outside the tumor using RADPOS 4D dosimetry system. The phantom was irradiated on an Elekta Infinity linac in both stationary and moving states. The dose delivery was simulated using delivery log files and the phantom motion recorded with RADPOS.

Results: Reproducibility of the phantom motion was determined to be within 1 mm. The phantom motion presented realistic features like hysteresis. MC calculations and measurements agreed within 2% at the center of tumor. Outside the tumor agreements were better than 5% which were within the positional/dose reading uncertainties at the measurement points. More than 94% of dose points from MC simulations agreed within 2%/2 mm compared to film measurements.

Conclusion: The deformable lung phantom presented realistic and reproducible motion characteristics and its use for verification of 4D dose calculations was demonstrated. Our 4DMC method is capable of accurate calculations of the realistic dose delivered to a moving and deforming anatomy during static and dynamic beam delivery techniques.

Keywords: Monte Carlo, deformable lung phantom, 4D dose calculation, VMAT

1. Introduction

Respiratory motion, as a type of intra-fraction motion, is one of the potential sources of deviations between the planned and delivered doses to tumors in the thorax and abdomen. Such motions and deformations can cause the blurring of the dose distribution, localized deformations of the dose distribution and interplay between motion of the target volume and multileaf collimator (MLC) [1–3].

Four-dimensional (4D) dose calculation algorithms can be used to account for the aforementioned impacts of the respiratory motion on the delivered dose distribution. In principle, 4D dose calculation algorithms calculate the dose on multiple instances of a continuously moving and deforming anatomy (e.g. different phases of the respiratory cycle) and map the cumulative dose to the reference (planning) phase of the anatomy [4–7]. Such dose mapping requires deformation vector fields (DVF\textsuperscript{s}) generated by a deformable image registration (DIR) algorithm. Different deformable image registration algorithms such as spline techniques might be used [8,9]. However, regardless of the registration...
technique used, the quality assurance of the DIR to ensure accuracy of DVFs is an important step, since accuracy of DVFs impacts the accuracy of dose calculations. Visual evaluation of the DIR is a preliminary and common test [8]. There are also quantitative methods including landmark tracking [8,10–14], contour comparison [8,13] as well as benchmarking against known deformations generated by a computational or physical phantom [8,11,15].

One advantage of 4D dose calculation algorithms is that they account for both translations and deformations of the anatomy that occur during respiration [4,5,16]. However, verification of any such algorithm requires a deformable phantom that is reproducible in motion/deformation as well as setup. In addition, it needs to mimic the physical and physiological characteristics of the human anatomy such as lung. Several research groups have developed tissue-equivalent phantoms that are capable of simulating motions and deformations induced by respiration. Kashani et al. [17] used a thoracic phantom as the exterior shell of a deformable lung phantom with an iodine-infused foam as lung. They inserted rigid tumors made from different materials in the foam and compressed/decompressed the foam using a programmable actuator system. The phantom was used as a validation tool for image registration algorithms [18]. Recently, Cheung et al. [19] constructed an externally and internally deformable phantom and tested its geometrical accuracy and reproducibility under sinusoidal and irregular motion patterns. They used the outer shell of a commercially available lung phantom with flexible anterior surface and filled it with a block of latex foam which was deformed using a programmable platform. Other groups [20,21] developed and validated real-tissue phantoms. Szegedi et al. [20] placed porcine liver between two rigid supporting membranes and induced the respiratory motion by pulling and pushing one membrane while the other was fixed. To evaluate performance of 4D magnetic resonance imaging (MRI), Remmert et al. [21] developed a dynamic phantom based on fresh porcine heart-lung organ placed in a chest wall made of a double-walled container filled with saline solution. The inflations and deflations of a water-filled balloon simulated the diaphragm while a vacuum pump connected to the container allowed expansion of the lung. Some recently developed phantoms [22–24] placed silicon or latex balloon filled with dampened sponge simulating lung with [23,24] or without [22] moving internal targets inside cylindrical bodies. One disadvantage of phantoms made of dampened sponge is their limitation to reproduce same structure setup due to the constant need to uniformly dampen the sponge before every measurement. Design of the mentioned phantoms did not allow dosimetric evaluations (to hold film or other dosimeters) and they were mainly validated geometrically.

Other groups designed phantoms for dosimetric validation purposes. Nioutsikou et al. [25] built a breathing phantom composed of two accordion-type flexible bottles mounted inside a container to simulate lungs. They also used dampened sponge to fill the lungs and placed a solid cylindrical tumor with several cuts to hold film in one of the lungs. Their phantom was used to quantify the impact of respiratory motion on the dose delivered to the tumor during static conformal and intensity-modulated radiotherapy (IMRT) plan deliveries. The phantom designed by Serban et al. [10] used a Lucite cylinder as external body filled with water and a latex balloon stuffed with dampened natural sponge to simulate lung. A deformable Dermasol tumor was incised (to hold radiochromic film and micro-MOSFET) and embedded in the sponge along with nylon wires and Lucite beads to emulate lung structures. Cherpak et al. [26] combined this phantom with a 4D dosimetry system, RADPOS [27] to verify dose and position in a deformable phantom. Court et al. [28] created a flexible resin tumor based on computed tomography (CT) images of an actual tumor and placed it inside an anthropomorphic breathing phantom. They inserted twenty micro-MOSFETs inside the tumor and
measured the effect of real-patient respiratory motion on the tumor dose during conformal, IMRT and volumetric-modulated arc therapy (VMAT) beam deliveries. Recently Mann et al. [29] developed a phantom made of porcine lung placed inside a PMMA cylindrical shell and inserted a 3D polymer gel dosimetry tumor at the center of the phantom. Evaluation of the phantom was performed by comparing measured and planned dose distributions from stationary and moving (with and without motion compensation) conditions of the phantom during static conformal beam deliveries.

Other research groups have performed experimental verification of dose accumulation algorithms using deformable phantoms [14,30–32]. Vinogradskiy et al. [31] used a deformable phantom made from slices of sponge and foam placed inside a cylinder and compressed/decompressed it with a piston attached to a programmable actuator to verify their 4D dose calculation algorithm implemented in a research version of Pinnacle³ treatment planning system (Philips Radiation Oncology System, Fitchburg, WI, USA). A cylindrical rigid target which contained one thermoluminescent dosimeter (TLD) and space for film in the sagittal and coronal planes was embedded in one of the sponges. Also, three films in plastic cassettes were sandwiched between the sponges in transverse plane. They compared measured and calculated dose distributions for sinusoidal and irregular motion patterns during conformal and IMRT beam deliveries and reported an average passing rate of 85% for the 5%/3 mm gamma criterion. In a study conducted by Niu et al. [32] a gel dosimeter was compressed with an actuation device. The phantom was irradiated with a conformal plan while being compressed at different amplitudes and the resultant dose distributions were compared against 3D dose accumulation calculations using a previously developed dose accumulation algorithm [33]. The average passing rate of 2%/2 mm gamma comparison was reported to be 92.7% while it was increased to 96.9% when the criterion was relaxed to 3%/3 mm. Another group [14] recently published their results on experimental verifications of a 3D Monte Carlo dose accumulation algorithm using mass-and-energy-congruent mapping method [6]. They designed a phantom made of sponge insert and tissue-equivalent tumor composed of bolus material and irradiated it with a static beam in stationary and deformed modes. Dose was measured with TLDs placed at multiple locations in the phantom. The DVFs required for dose accumulation were generated using the VelocityAI (Varian Medical Systems, Palo Alto, CA, USA) and Elastix [34,35] software packages. Total accumulated dose with the two sets of DVFs was observed to agree within 5.7% of the TLD measurements. This was slightly higher than the 5% clinical acceptance.

In our previously published work [16], we presented the experimental validation of our novel 4D Monte Carlo dose calculation method, based on the voxel warping approach [7], to reconstruct the dose delivered to a rigidly moving respiratory motion phantom (Quasar, Modus medical, London, ON, Canada). To evaluate performance of our 4DMC tool in a more realistic, continuously deforming anatomy, we developed a novel deformable phantom with characteristics similar to lung, containing a tissue-like elastic tumor. This phantom is reproducible in motion/deformation as well as structure setups. Moreover, design of the phantom allows dosimeters (film and RADPOS [27]) to be inserted and removed easily and thus enables measurement of point dose and dose distribution. Placing RADPOS inside the phantom enables us to verify the internal motion and obtain motion trace for 4DMC simulations. Dose deliveries with the deformable phantom were performed on an Elekta Infinity linac with Agility MLCs for static and VMAT beam deliveries and delivery log files were recorded. Simulated dose distributions and point dose values were then compared against measurements to evaluate the accuracy of simulations.
2. Materials and methods

2.1. Deformable phantom design and construction

A deformable lung phantom (Fig. 1a) was developed based on modifications to the design previously proposed by Serban et al. [10]. To simulate the lung, flexible foam with density of $0.17 \pm 0.02 \text{ g/cm}^3$ was molded into a cylindrical insert which was placed inside an external cylindrical body made of Lucite. The foam composition was selected to have similar CT numbers (-830 $\pm$ 37) HU (Hounsfield Units) to human lung tissue. A plug (Fig. 1b) was molded from the same material and was fixed into a cylindrical hollow inside the lung insert. The plug was used to hold a 2.6 cm diameter tumor made of silicon rubber with density of $1.05 \pm 0.04 \text{ g/cm}^3$ (Fig. 2a). 32 Lucite beads, with diameter of 1.6 mm, were injected as landmarks throughout the phantom to help with the deformable registration and quantifying the phantom motion (Fig. 2b). At the inferior end of the phantom, a piston attached to a DC motor simulated the diaphragm and the breathing motion by compressing and decompressing the lung phantom in the S-I direction, according to a sinusoidal motion with 4 cm peak-to-peak (P-P) respiratory motion with varying motion periods.

![Fig. 1](image1.png)

**Fig. 1.** (a)The deformable lung phantom consisting of the flexible foam, plug, piston and motor, and (b) Cylindrical plug containing the tumor.

![Fig. 2](image2.png)

**Fig. 2.** (a) Silicon rubber tumor inside the plug and (b) Lucite beads injected throughout the phantom.
2.2. Motion reproducibility

To assess the motion reproducibility of the phantom in 3D (SI, LR and AP), the RADPOS 4D dosimetry system [27] was used. RADPOS probe consists of a microMOSFET dosimeter and an electromagnetic positioning system for real-time tracking purposes. The reported uncertainty in the root-mean-square (RMS) displacement of RADPOS is ± 0.21 mm [27]. RADPOS detectors were placed at the tumor center (approximately 9 cm from piston) as well as on the top and bottom surfaces of the plug (Fig. 3a,b,c). Motion of the tumor and lung were recorded at a temporal resolution of 100 ms when a sinusoidal motion with a 3.3 sec period was applied. The RADPOS on the bottom surface of the plug was aligned with the one inside the tumor while the top surface RADPOS was offset approximately 1.5 cm from the tumor center along the SI axis. Motion measurements were repeated 10 times independently to evaluate the motion reproducibility of the phantom. Displacement magnitude for these points in the SI direction was also measured on the 4DCT scans of the phantom using End-of-Inhale (EOI) and End-of-Exhale (EOE) images.

2.3. 4DCT image acquisition

4DCT images of the phantom were acquired using a helical CT scanner (Brilliance CT Big Bore, Philips, Amsterdam, the Netherlands) at an image resolution of 0.05×0.05×0.2 cm³ resulting in an image matrix of 512×512×226 voxels. Pitch values and the gantry rotation times for the scans were 0.1 and 0.5 sec, respectively.

2.4. Image registration

Deformable image registration (DIR) of the 4DCT images was performed in VelocityAI 3.2.0. The algorithm utilized by Velocity for deformable registration is a modified B-spline algorithm [9] which uses mutual information to evaluate image matching. This combination provides an accurate, stable and fast algorithm for deformable registration of CT images [9].

The EOI and EOE 4DCT scans were chosen as primary (reference) and secondary volumes, respectively. The beads as well as the tumor were contoured as structures in both images and the structure-guided deformable registration in
Velocity was used to deformably register the EOE to the EOI image. Velocity outputs the deformation vectors at the center of mass of each voxel in the coordinate system of the primary image, pointing to the secondary. The deformation vector fields (DVFs) were exported from Velocity. To evaluate the accuracy of registrations, target registration errors for each landmark were calculated in Velocity by identifying the matching bead locations in the primary and secondary images.

2.5. Treatment plans

Two treatment plans for 6 MV photon beams from Elekta Infinity linac with Agility MLCs (Elekta AB., Stockholm, Sweden) were created on the EOI 4DCT scans of the phantom in Monaco V.5.11.01 (Elekta AB., Stockholm, Sweden). A gross target volume (GTV) was created by contouring the tumor with no margins added to compensate for motion. Both plans were designed to deliver 100 cGy to the center of the tumor.

The first treatment plan was a static 3×3 cm² field and the second was a VMAT plan with 67 control points delivering a full arc starting and ending at 180° with angular spacing of 6.6° between control points. The dose calculation algorithm used by Monaco is XVMC (X-ray Voxel Monte Carlo) [36]. In order to be consistent with measurements (film and RADPOS calibrated in water), dose to water (Dw) was calculated in Monaco using a 2 mm dose calculation grid to achieve a statistical uncertainty of 1%. Dose distributions corresponding to the static and VMAT plans on the EOI image are shown in Fig. 4 and 5, respectively. Static and VMAT plans covered the GTV with the 85% and 90% isodose lines, respectively.

![Fig. 4. Static 3×3 cm² square plan: dose distribution on (a) coronal, (b) sagittal and (c) axial planes.](image-url)
2.6. Dose measurements: Film and RADPOS

Dose measurements inside the plug were performed using calibrated Gafchromic film (EBT3, Ashland, Wayne, NJ, USA) and RADPOS as shown in Fig. 6. To measure point dose outside the plug (inside the lung), RADPOS detectors mounted on top and bottom plug surfaces (Section 2.2) were utilized. Details on dose measurements with RADPOS can be found in previously published works [16,27].

The RADPOS probe inside the plug was fixed in a groove that was embedded inside the plug as well as the tumor during the molding process so that the point of measurement of RADPOS probe was located at the center of the tumor. Film, as shown in Fig. 6 was taped on top of the RADPOS probe.

The EBT3 film and RADPOS detectors were cross-calibrated against an A1SL ionization chamber (Standard imaging Inc., Middleton, WI, USA) using a 6 MV photon beam and 10×10 cm² field size. The calibration process is described in detail in our previous work [16]. The total dosimetric uncertainty of RADPOS measurements were determined to be approximately 1.4%, 1.9% and 1.7% for the detectors at the tumor center, top and bottom surfaces of the plug, respectively. This includes the uncertainties in the dosimeter calibration and beam delivery conditions. The
The main contributors to the dosimeter calibration uncertainty included uncertainties corresponding to IC calibration ($N_{D,w}$ from clinical data $\sim 1.0\%$, $k_0 \sim 0.4$ [37]), IC readings (0.2%, 0.0%, 0.0%), RADPOS reproducibility (0.5%, 1.4%, 1.0%), and Solid Water phantom material variability [38]. The uncertainty of the beam delivery conditions was determined to be 0.3% including uncertainties corresponding to SSD, depth and field size settings, as well as temperature and pressure correction factors [37]. For film measurements, the total dosimetric uncertainty was determined to be approximately 2.3%. Scanner uniformity, lateral correction, film inhomogeneity, energy and angular dependence, calibration curve fit accuracy (net OD to dose) and intra-batch variations were the main contributions of uncertainty associated with EBT3 films (1.8%) [39]. Total uncertainty of the film calibration (1.3%) was determined based on parameters such as IC calibration (same as for RADPOS) and Solid Water phantom material variability [38]. The uncertainty related to beam delivery (0.3%) was determined to be similar for film and RADPOS measurements. Total dosimetric uncertainties of film and RADPOS measurements were obtained by quadrature sum of all contributing sources of uncertainty.

2.7. Irradiations of deformable lung phantom

Both treatment plans from Section 2.5 were exported into the Elekta MOSAIQ RadOnc system and delivered to the lung phantom on an Infinity linac with Agility MLCs. For the 3x3 cm$^2$ square plan, 107.4 MU was delivered at a nominal dose rate of 600 cGy/MU. The VMAT plan delivered 111.5 MU at a varying dose rate. To extract information on the delivered plans including cumulative MU, gantry, MLC, jaws and table positions, delivery log files were recorded to be used for MC simulations. These log files have a temporal resolution of 40 ms.

Two sets of irradiation were performed on stationary (no motion) and three sets on breathing (sinusoidal motion: 3.3 sec period, 4 cm P-P motion of piston) states of the phantom for each plan delivery. The phantom was set up so that the center of the tumor (as marked on the phantom) was aligned with the beam isocenter. Film and one RADPOS detector were placed inside the plug and two RADPOS detectors were mounted on the top and bottom surfaces of the plug. Film was used to measure the dose profiles along the phantom motion while point doses were measured with RADPOS detectors. During deliveries to the moving phantom, the motion was recorded with RADPOS at a temporal resolution of 100 ms. Prior to irradiation, the computer clock times of both the linac and RADPOS computers were synchronized. This was necessary to ensure synchronization of the beam delivery data in the log file with the phantom motion recorded with RADPOS. This synchronization is good within ms.

2.8. Monte Carlo user codes and simulation parameters

All Monte Carlo simulations were performed using EGSnrc [40] (V4-2.4.0, National Research Council of Canada, Ottawa, ON, Canada). The BEAMnrc user code [41] was used to model the Elekta Infinity linear accelerator with Agility MLC. Details on this model including its parameter and validation have been presented in the previously published work [16]. Dose calculations in the lung phantom were performed using the DOSXYZnrc [42] and 4DdefDOSXYZnrc [16] user codes on stationary and deforming states, respectively. Calculated dose from Monte Carlo simulations was converted to absolute dose using the following formulation:
\[
D \text{ (cGy)} = \left(\frac{D}{\# \text{ of incident particles}}\right)_{\text{MC individual simulation}} \times \frac{1 \text{ cGy}}{\text{MU}} \times \text{MU}_{\text{del}}
\]

where, \( \text{MU}_{\text{del}} \) is the monitor units (MU) delivered by a linear accelerator. In this formula \( \frac{D}{\# \text{ of incident particles}} \) represents the dose scored per number of incident particles in Monte Carlo simulations. The calibration simulation was performed in water for a square field of 10\( \times \)10 cm\(^2\) and SSD of 100 cm and dose was scored at a depth of 10 cm.

An in-house Python script was written to generate input files required for MC simulations from the linac delivery log files. The photon cutoff energy (PCUT) and electron cutoff energy (ECUT) were set to 0.01 and 0.7 MeV, respectively, and the electron range rejection was set to 2 MeV for all simulations. Dose calculations were performed with 1.5\( \times \)10\(^8\) and 3.0\( \times \)10\(^8\) histories on the stationary and deforming geometries, respectively, to achieve a mean relative statistical uncertainty [43] of 0.4\% over all voxels with doses greater than 50\% of the maximum dose. Achieving this statistical uncertainty required 12-15 and 31-37 CPU core hours per calculations on the stationary and deforming geometries, respectively. All simulations were performed on the Carleton University Research Computer Cluster which consists of 648 processing cores with 2, 4, and 12 core Intel Xeon CPUs at 2.50-3.00 GHz frequency.

2.9. Dose calculation geometry

To generate the voxelized dose calculation geometry (egsphant file), 4DCT scans of the EOI were resampled to a resolution of 0.2\( \times \)0.2\( \times \)0.2 cm\(^3\). Voxel information was derived from CT data and densities were assigned according to the HU/density calibration data from the Big Bore CT scanner using the method introduced by Seco et al. [44]. This approach allowed direct calculation of \( D_w \) in MC simulations to be compared against measurements with film and RADPOS.

2.10. Monte Carlo simulations: stationary and deforming phantom

For simulations of the 3\( \times \)3 cm\(^2\) field and VMAT plan on the stationary state of the phantom, particle sources 9 [42] and 21 [45] in DOSXYZnrc were used, respectively. They both use a full BEAMnrc linac model which eliminates the need to store phase space data separately. Source 21 allows synchronization between the motion of linac components and dose calculation geometry. Details on how 4DdefDOSXYZnrc performs 4D simulations on a continuously deforming anatomy are presented in our previously published work [16].
2.11. Comparison metrics

For each set of irradiations, dose profiles along the motion path of phantom (SI direction) were compared from MC simulations against film measurements. Due to the deformations in the phantom, it was only possible to measure dose profiles partially with film. To match the dose grid resolution of film with the one from MC simulations (2 mm), a moving average filter was applied to the film readings. For comparison purposes, a 1D gamma analysis [46] with a 2% dose-difference and 2 mm distance-to-agreement criterion was utilized with film dose used as the reference. Point dose comparisons were performed between MC simulations and measurements at the center of the tumor (film and RADPOS) as well as top and bottom surfaces of the plug (RADPOS). These comparisons are quoted as percentage of the measured dose value at the point of comparison.

3. Results

3.1. Deformable phantom performance

The density and volume of lung insert were found to increase by 11.5% and decrease by 28.7%, respectively, from EOI to EOE. Various levels of hysteresis (Fig. 8) were observed based on trajectories of selected landmarks (Lucite beads), with larger amounts in points closer to the piston and smaller in points further away. This shows that the lung insert follows different deformation paths between compression and decompression.
3.2. Assessment of motion reproducibility

Measured motion magnitudes in the AP, LR and SI directions along with their reproducibility for the points described in Section 2.2 are shown in Table 1. The maximum range of motion, as expected, is along the SI direction. The SI displacements measured using 4DCT scans shown in the last column of the table are accurate within the resolution of CT image slices (2 mm) and agree with the range measured with RADPOS detectors. The AP and LR motions are less than 1 mm because the points of measurements were approximately at the center of the phantom.
Table 1

<table>
<thead>
<tr>
<th>Measurement point</th>
<th>AP (cm)</th>
<th>LR (cm)</th>
<th>SI (cm)</th>
<th>SI from 4DCT (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tumor Center</td>
<td>0.06 ± 0.01</td>
<td>0.04 ± 0.01</td>
<td>1.37 ± 0.06</td>
<td>1.46</td>
</tr>
<tr>
<td>Top surface</td>
<td>0.10 ± 0.01</td>
<td>0.04 ± 0.02</td>
<td>1.23 ± 0.09</td>
<td>1.17</td>
</tr>
<tr>
<td>Bottom surface</td>
<td>0.07 ± 0.00</td>
<td>0.03 ± 0.01</td>
<td>1.29 ± 0.07</td>
<td>1.25</td>
</tr>
</tbody>
</table>

3.3. Validation of image registration

The registration accuracy was found to be 0.05 ± 0.03, 0.04 ± 0.03 and 0.06 ± 0.05 cm in the AP, LR and SI directions, respectively. The overall 3D registration error was found to have an average value of 0.1 ± 0.04 cm with minimum and maximum error values of 0.01 and 0.17 cm. The maximum error value was found to be within the resolution of the CT image slices (0.2 cm) which was the limiting factor to achieve a more accurate registration.

Performance of the image registration algorithm can also be evaluated visually by showing overlaid images of the EOE and EOI before and after registration (Fig 9.a,b). A difference map of the deformed EOE image from the primary image (EOI) as shown in Fig. 9c. Regions in light gray present the lowest differences between the deformed and primary images.

![Image](image_url)

Fig. 9. Visual evaluation of the deformable image registration on coronal view: (a) non-deformed EOE (gray) overlaid on EOI (pink), (b) deformed EOE overlaid on EOI and (c) deformed EOE subtracted from EOI.
3.4. Stationary phantom comparisons

Figure 10 shows an example of dose profiles from MC simulations and film measurements for the 3×3 cm² and VMAT plan deliveries on the stationary phantom.

![Dose profiles comparison](image)

**Fig. 10.** Comparison of dose profiles for (a) 3×3 cm² and (b) VMAT beam deliveries on the stationary phantom along the SI direction.

Very good agreement was observed between measured and simulated dose profiles. All dose points from MC simulations passed a 2%/2 mm gamma comparison for the 3×3 cm² square field. For the VMAT plan, this value was found to be 97.8%. The average gamma value from two sets of deliveries were calculated to be (98.2±2.5)% and (98.9±1.6)% for stationary and deforming phantoms, respectively. Dose values measured (film and RADPOS) and simulated at the center of the tumor (from Fig 10) are shown in Table 2. The reported uncertainties are experimental and statistical uncertainties corresponding to measurements and MC simulations, respectively.

**Table 2**

<table>
<thead>
<tr>
<th>Plan</th>
<th>Dose (cGy)</th>
<th>Measurements</th>
<th>MC</th>
<th>Measurements</th>
<th>Film</th>
<th>RADPOS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Static 3×3 cm²</td>
<td>99.3±0.4%</td>
<td>99.9±2.3%</td>
<td>99.7±1.4%</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>VMAT</td>
<td>99.8±0.4%</td>
<td>101.7±2.3%</td>
<td></td>
<td>99.7±1.4%</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

All dose values from MC simulations were found to be approximately within 2% of film and RADPOS measurements for both plan deliveries. Comparisons of both irradiation sets also showed an average agreement of 2% or better. Table 3 shows dose values for point dose measurements at the top and bottom surfaces of the plug with their corresponding statistical and experimental uncertainties.
Table 3
Dose values at the top and bottom surfaces of the plug from MC simulations and RADPOS measurements on the stationary phantom. The top surface RADPOS was placed at an offset position of 1.5 cm from the tumor center.

<table>
<thead>
<tr>
<th>Plan</th>
<th>Top surface</th>
<th>Bottom surface</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MC</td>
<td>RADPOS</td>
</tr>
<tr>
<td>Static 3×3 cm²</td>
<td>19.8±0.4%</td>
<td>20.7±1.9%</td>
</tr>
<tr>
<td>VMAT</td>
<td>63.6±0.4%</td>
<td>63.2±1.9%</td>
</tr>
</tbody>
</table>

For the measurement point at the bottom surface of the plug, dose values from RADPOS measurements and MC simulations agreed within 1% of each other. At this dose point, the dose reading/positional uncertainty was calculated to be less than 3% for both plan deliveries. Agreements on the top surface of the plug were found to be better than 5%. Considering that this point is positioned in high dose gradient and low dose region (penumbra) of the profile, the uncertainty on the dose reading or positional uncertainties were calculated to be over 20 and 10% for the static and VMAT beam deliveries, respectively. The larger uncertainties of the static plan associate with the steeper penumbral of dose profiles. Agreement levels were observed to be better than 1 and 5% for bottom and top plug surface dose points when two sets of irradiations were included.

3.5. Moving and deforming phantom comparisons

Examples of dose profiles from MC simulations and film measurements for the deforming phantom are shown in Fig. 11 for both plan deliveries.

![Fig. 11. Comparison of dose profiles for (a) 3×3 cm² and (b) VMAT beam deliveries on the deforming phantom along the SI direction.](image)

As it can be seen in Fig. 11, there is a good agreement between simulated and measured dose profiles on the deforming phantom for both plan deliveries. All dose points from MC simulations passed the 2%/2 mm gamma...
comparison for the VMAT delivery. As for the 3×3 cm² square field delivery, this value was better than 93%. Agreement levels when considering all three sets of irradiations were found to be at (98.1±3.2)% and (94.6±1.4)% for VMAT and static plan deliveries, respectively. The slightly worse agreement for the static beam is a result of higher dose values from film at the center of the profile as can be seen from Fig 9a. Table 4 shows dose values at the center of the tumor.

**Table 4**
Dose values at the center of the tumor from RAPDOS, film and MC simulations for static and VMAT beam deliveries on the deforming phantom.

<table>
<thead>
<tr>
<th>Dose (cGy)</th>
<th>Plan</th>
<th>MC</th>
<th>Measurements</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Film</td>
<td>RADPOS</td>
</tr>
<tr>
<td>Static 3×3 cm²</td>
<td>87.9±0.4%</td>
<td>87.3±2.3%</td>
<td>87.3±1.4%</td>
</tr>
<tr>
<td>VMAT</td>
<td>97.8±0.4%</td>
<td>97.7±2.3%</td>
<td>98.3±1.4%</td>
</tr>
</tbody>
</table>

As we can see from values shown in Table 4, MC simulations and measurements were found to have an agreement of better than 1%. Overall and for all three sets of irradiations, agreements were found to be within 2% for both plans. In Table 5, point doses at the top and bottom plug surfaces are reported as well.

**Table 5**
Dose values at the top and bottom surfaces of the plug from MC simulations and RADPOS measurements on the deforming phantom. The top surface RADPOS was placed at an offset position of 1.5 cm from the tumor center.

<table>
<thead>
<tr>
<th>Dose (cGy)</th>
<th>Plan</th>
<th>Top surface</th>
<th>Bottom surface</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>MC</td>
<td>RADPOS</td>
</tr>
<tr>
<td>Static 3×3 cm²</td>
<td>13.5±0.4%</td>
<td>13.4±1.9%</td>
<td>74.8±0.4%</td>
</tr>
<tr>
<td>VMAT</td>
<td>30.1 ±0.4%</td>
<td>30.4±1.9%</td>
<td>77.5±0.4%</td>
</tr>
</tbody>
</table>

Overall, the agreements were found to be slightly over 4% between MC simulations and point dose measurements using RADPOS at the bottom surface of the plug. This value complies with the positional/reading uncertainties corresponding to the dose values at this point of measurement which are approximately 7 and 4% for the 3×3 cm² square field and VMAT deliveries, respectively. At the top surface, the agreements were better than 1% while the dose reading/positional uncertainties were larger than 20 and 15% for static and VMAT deliveries, respectively. Similar to the
stationary phantom measurements, the larger uncertainties of the static plan are a result of the steeper penumbras of dose profiles. Agreement levels of point dose values were found to better than 5% when all sets of irradiations were taken into account.

4. Discussion

Validation of 4D Monte Carlo simulations, using the 4DdefDOSXYZnrc user code to calculate the dose delivered to a moving and deforming lung phantom was the main objective of this study. For these verifications, a deformable lung phantom composed of compressible tissue-equivalent foam containing a moving non-rigid tumor was constructed and characterized. The foam was compressed/decompressed with a piston driven by a DC motor. Design of the phantom allowed inclusion/removal of film and RADPOS for dose and position measurements. The phantom showed good motion reproducibility (~ 1 mm) in 3D as well as hysteresis features that were extracted through landmarks injected throughout the phantom. Deformable registration of the EOE to EOI CT images of the phantom also showed a good level of accuracy which stayed within the image resolution of the CT dataset, i.e. 2 mm.

Static and VMAT plans were delivered to the phantom in stationary and deforming states. Point dose measurements with RADPOS were performed inside and outside the tumor (still inside the lung). In addition, film was used to measure dose distributions along the motion path of the phantom (SI). Recorded displacements with RADPOS as well as the DVFs generated by deformable registration of the phantom with VelocityAI were used as input to 4DMC simulations to model the phantom motion and deformation. Our results show that the average agreement of point dose values at the center of the GTV (i.e. tumor) from MC simulations and measurements by film and RADPOS are within 2%. As for the dose points outside the tumor (i.e. top and bottom surfaces of the plug), simulations and measurements were found to have an average agreement of 5% or better. These agreements were found to be within the calculated positional uncertainties of these dose points. The agreements between the simulated and measured dose profiles, extracted along the SI direction (motion path of the phantom), were good as well. Gamma comparisons of 2%/2 mm showed an overall passing rate of better than 94%. Over 98% of dose points passed the Gamma comparison when relaxed to 3%/3 mm criterion.

The accuracy level of DVFs used in 4D simulations could affect the accuracy of simulations. The DVFs used in our work were found to have accuracy better than 1.7 mm in 3D trajectories. RMS differences (in 3D) between average measured motions and calculated DVFs at the tumor center as well as top and bottom surfaces of the plug were found to be 0.9, 0.6 and 1 mm, respectively, which were better than the worst accuracy within the DIR. The hysteresis effect of the phantom was not included in simulations and EOI to EOE motion was modeled by linear interpolation. This approximation could be a potential source of uncertainty if the level of hysteresis was large.

The impacts of larger/smaller motion amplitudes on dose values were investigated by modifying the DVFs. Since the largest motion corresponds to SI direction, change in the dose due to change in this component was studied. The amplitude was changed by 4% (higher and lower) which converted to about ±0.6 and ±0.5 mm at the tumor center and top and bottom plug surfaces, respectively. These values compensate approximately for the variations between motion amplitudes measured with RADPOS and modeled by DVFs. The impact on static and VMAT plans and for each dose
measurement point was studied separately. For the static plan, decrease of approximately 4% in dose was observed at the center of tumor when larger amplitudes were used. The reason for such deviations is the fact that this point dose leaves the radiation field during the phantom motion/compression. For smaller amplitude of motion dose was increased only by 1-2% since dose point still remains in the radiation field. Looking into similar values for the VMAT plan, dose increased/decreased by 1-2%. As reported values suggest, dose deviations for the VMAT plan are not as large as for the static plan. The main reason is the wider radiation field and as a result, lower dose gradient of the VMAT compared to the static plan. For both plan deliveries, at the bottom surface of the plug, dose was deviated by 1-2% due to the change in the motion amplitude. Dose deviations at the point dose at top surface of the plug, were observed to be over 4% for both plans because it was originally in a high dose gradient region. Like positional uncertainties, DVF errors and differences in the simulated vs. measured amplitudes are other sources of uncertainty that can account for differences observed between calculated and measured dose values.

The difference in temporal resolutions of RADPOS (100 ms) and log files (40 ms) could be one potential source of experimental uncertainty as it impacts the accuracy of the detecting starting phase of respiratory cycle. These system delays convert to positional inaccuracies of about 0.5 and 0.4 in the SI direction, at the tumor center and top and bottom plug surfaces, respectively. Moreover, the PC clock synchronization between linac and RADPOS computers could be another source of uncertainty to determine the starting phase of motion. In addition, we have the position measurement uncertainty of RADPOS which is better than 0.2 mm. After investigation, it was observed that such positional uncertainties can cause point dose deviations of approximately 1-2% in the target volume as well as bottom surface of the plug. Dose value at the top surface of the plug is more sensitive to such positional uncertainties as a result of being placed in high dose gradient region. Thus, dose deviations of as large as 7-8% might be observed at this point. Accurate detection of the starting phase of the respiratory cycles becomes more essential when delivering highly modulated plans (due to the interplay effect) or plan with shorter delivery times.

Another potential source of uncertainty in MC simulations of VMAT deliveries corresponds to the interpolation between control points performed by source 21. However, these control points are defined according to delivery log files with a high sampling rate of 40 ms. Analyzing delivery log files revealed that this sampling rate corresponds to maximum change of 0.2° and 0.1 mm in the gantry angle and MLC positions between two consecutive interpolation points, respectively. With respect to MU fraction, this change does not exceed 0.001. Thus, high sampling rate of log files help minimizing the potential errors caused by interpolations between control points.

5. Conclusions

A novel reproducible deformable lung phantom with capability to hold film and RADPOS was developed and validated. We investigated and established the accuracy of 4D Monte Carlo simulations, using the 4DdefDOSXYZnrc user code, of dose delivered to the phantom during sinusoidal motion with amplitude of ~1.4 cm and period of 3.3 sec. Measurements were performed on an Elekta Infinity linac equipped with Agility MLCs during static 3×3 cm² square and VMAT plan deliveries. Delivery log files were used to reproduce the measurements during these deliveries. Agreement of the dose values were approximately 2 and 5% inside and outside the target volume, respectively. The results of this
work complement findings from our previous study performed in a rigidly moving phantom [16]. Our recent findings demonstrate that combining position tracing with RADPOS and DVF s generated by a reliable DIR algorithm (to model motion and deformation of the anatomy) with our 4DMC simulations, leads to accurate dose calculations in a deforming anatomy. Our future ongoing work is to modify the phantom and verify the accuracy of our 4DMC tool in the presence of irregular respiratory motion patterns. The methodology established in this work can be extended to measurements/simulations with other linacs (e.g. Varian) that allow recording delivery log files.

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References


