Journal of Physics: Conference Series **102** (2008) 012008 doi:10.1088/1742-6596/102/1/012008

A 4D treatment planning tool for the evaluation of motion effects on lung cancer treatments

M Ding[1], F Newman[1], L Gaspar, B Kavanagh1, K Stuhr[1], D Raben[1], J S Li[2] and C-M Ma[2]

[1]Radiation Oncology Department, University of Colorado at Denver and Health Science Center, Aurora, CO 80010, USA
[2]Radiation Oncology Department, Fox Chase Cancer Center, Philadelphia, PA 19111, USA

E-mail: meisong.ding@uchsc.edu

Abstract. In this study, a 4D treatment planning tool using an analytical model accounting for breathing motion is investigated to evaluate the motion effect on delivered dose for lung cancer treatments with three-dimensional conformal radiotherapy (3DCRT). The Monte Carlo EGS4/MCDOSE user code is used in the treatment planning dose calculation, and the patient CT data are converted into respective patient geometry files for Monte Carlo dose calculation. The model interpolates CT images at different phases of the breathing cycle from patient CT scans taken at end inspiration and end expiration phases and the chest wall position. Correlation between the voxels in a reference CT dataset and the voxels in the interpolated CT datasets at any breathing phases is established so that the dose to a voxel can be accumulated through the entire breathing cycle. Simulated lung tumors at different locations are used to demonstrate our model in 3DCRT for lung cancer treatments. We demonstrated the use of a 4D treatment planning tool in evaluating the breathing motion effect on delivered dose for different planning margins. Further studies are being conducted to use this tool to study the lung motion effect through large-scale analysis and to implement this useful tool for treatment planning dose calculation and plan evaluation for 4D radiotherapy.

1. Introduction

Four-dimensional (4D) radiotherapy was put forward at the ASTRO 2003 Panel Discussion of "Time ─ the 4th dimension in radiotherapy" [1]. The advantage of 4D radiotherapy is the explicit inclusion of the temporal changes in anatomy during the imaging, planning and delivery phases of radiotherapy. Several 4D CT scanners are commercially available and 4D treatment planning has been investigated [2-10] with in-house programs to perform the deformable image registration and to map the doses based on 4D CTs at different respiration phases to a reference CT. A feasibility study of 4D dose delivery using dynamic multileaf collimator-based respiratory motion tracking was also reported [11].

We have demonstrated a 4D Monte Carlo treatment planning system to study image registration and dose correlation to account for thoracic motion in radiation therapy of breast cancer [12], using an analytical model to interpolate CT-based anatomy between end inspiration and end expiration. The novelty of the model is that it accounts for breathing motion using a conventional CT scanner with a minimal number of CT scans. In this paper, we extend this model to evaluate the motion effect on the delivered dose for lung cancer treatments with three-dimensional conformal radiotherapy (3DCRT).

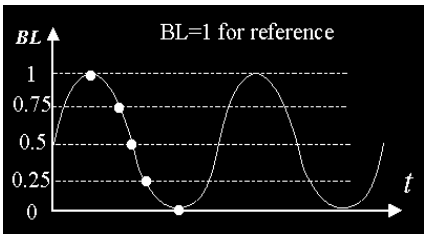


Figure 1. The sinusoidal breathing pattern used in this study. The x-axis is the time-scale and the y-axis is the normalized chest wall position, i.e., the relative breathing level (*BL*).

2. Methods and materials

An analytical model has been developed to interpolate a patient's CT image at any phases of a breathing cycle based on two initial CT scans taken at end inspiration and end expiration, and on the chest wall motion related to the patient's breathing phases. The 3D dose for the patient geometry built from the CT image at any breathing phases was calculated using Monte Carlo simulations with the EGS4/MCDOSE code [15]. Correlation between the voxels in the patient dose calculation geometry built from the end inspiration image and the voxels in the interpolated image at an individual breathing phase was established in order to accumulate the dose delivered to the same voxel from all the breathing phases.

In this study, we used an assumed chest wall motion pattern as indicated in figure 1. The x-axis is the time-scale and the y-axis is a relative value defined as the breathing level *(BL),* which represents the chest-wall position. If *BL* = 1 is defined for end inspiration and *BL* = 0 for end expiration, there is a corresponding *BL* value for any particular phase of a breathing cycle. The CT data at end inspiration and end expiration were obtained by coaching patients to hold their breath at these two extreme breathing phases. For Monte Carlo dose calculation the patient CT images at different breathing phases are converted to corresponding patient geometry files [13], which define the three-dimensional rectilinear voxel (volume element) geometries at these breathing phases. All the voxels in the patient simulation geometry have uniform dimensions. The material and density in a voxel can be determined from the CT numbers directly, or the electron density data derived from the patient CT data. There are two steps to determine the delivered dose accounting for the motion effect. The first step is to interpolate the CT image in any breathing phase of interest. The second step is to accumulate the dose distribution from the interpolated CT images to a reference CT image. The CT image at the end inspiration has been used as the reference image in this study.

2.1. Interpolation of the CT image

One of the features of our model is that it establishes the correspondence of voxels for different CT images at various breathing phases. The details have been given in a previous paper [12]. For completeness, here we summarize the process of image interpolation based on voxel correspondence in the x-y plane (axial plane) and slice correspondence in the z-direction (superior-inferior direction).

In the x-y plane (axial plane), first we find out the corresponding points between the external body contour points on the end inspiration CT image and on the end expiration CT image. The total length of each external contour is calculated. Then we divide both contours into the same number of sections with equal arc length, and list their middle points in the same order. The points (x1n, y1n) on the contour of the end inspiration CT image are then associated with those (x2n, y2n) on the contour of the end expiration CT image. Since we set BL=1 for the end inspiration phase and BL=0 for the end expiration phase, for an arbitrary breathing phase with a breathing level of BL, we have:

$$x_n = x2_n + BL \cdot \left(x1_n - x2_n\right) \quad (1)$$

$$y_n = y2_n + BL \cdot \left(y1_n - y2_n\right) \quad (2)$$

where (x_n, y_n) is the coordinate of the corresponding point on the contour of the interpolated CT image at the arbitrary breathing phase. Here we have assumed that the body will expand or compress proportionally based on the breathing level. Second we correlate each voxel in the interpolated CT image with that in the end inspiration CT image. For an arbitrary voxel in the interpolated CT image, there is a corresponding contour point (point ② in [12]) by extrapolating the straight line connecting the origin (a stationary point) and the center of this voxel. We calculate the ratio of the distance between point ② and the origin and the distance between the center of this voxel (voxel ①in [12]) and the origin. Using the equations (1-2), we establish the correspondence of point ② to the coordinates on the outer contour of the end inspiration CT image (point ③ in [12]). If we assume in the *x-y* plane the thoracic motion due to respiration only causes the CT image to change proportionally, the corresponding voxel (voxel ④ in [12]) in the end inspiration CT image can be determined by a linear interpolation. Therefore, voxel ④ will be located on the straight line connecting the origin and point ③. The ratio of the distance between point ③ and the origin and the distance between voxel ④ and the origin will be the same as the ratio for the interpolated CT image. Voxel ④ in the end inspiration CT image is therefore associated with voxel ① in the interpolated CT image.

In the z-direction we divide the CT image into three sections. Section I is below the lung volume, all the internal organs are assumed to shift by the same distance. The shift distance depends on the breathing level. Section II includes all the lung volume, the lung volume compresses/expands proportionally during respiration. Section III is above the lung volume, there is no organ movement in the z-direction during respiration.

2.2. Monte Carlo dose calculation

The Monte Carlo method is a statistical simulation method, which can accurately model the physical processes involved in radiation therapy and is powerful in dealing with any complex geometry. The EGS4/MCDOSE user code, developed by the Stanford group [13], was used for dose calculations in this study for 3DCRT. MCDOSE uses a multiple-source model to reconstruct the treatment beam phase space. Based on measured beam data acquired during accelerator commissioning, source-model parameters are adjusted through an automated procedure [14]. For 3DCRT dose calculation, the input data include the photon beam source file, the cross-section data file for different materials, and the 3D patient geometry file. The dose calculation geometry has a 128 x 128 x 128 grid with 3 mm voxels for dose scoring. MCDOSE runs on the Linux operating platform and it takes 1-4 hours CPU time on a 2 GHz Pentium 4 PC to compute the dose distribution for a typical 3DCRT plan.

2.3. Dose accumulation for a given breathing pattern

After calculating a sequence of 3D dose data for different interpolated CT images at the breathing levels of interest, we accumulated the dose to a reference CT image — the end inspiration CT image was chosen for this study. Using the voxel correspondence mentioned above, we accumulated the dose from the interpolated CT images at any phase (related to a specific breathing level) of a breathing pattern to the reference CT image. The time weight (i.e., the relative duration) for each breathing level may be different due to the shape of individual breathing patterns. The time weight is determined by the relative duration of the breathing level (corresponding to a specific phase) in the whole breathing cycle.

To compare the accumulated dose distribution taking into account the breathing motion effect with the normal planning dose distribution calculated using the reference CT image, the accumulated dose should be scaled by dividing the total accumulated time weights, which is the sum of the time weights at different breathing levels we have used. If we assume that voxel V_{ijk} (e.g., voxel ④ in section 2.1) in end inspiration CT image has a corresponding voxel $V_{i'j'k'}$ (e.g., voxel ① in section 2.1) in the

interpolated CT image for BL_m (m = 1, 2, …, N), the accumulated dose in voxel V_{ijk} can be expressed as

$$D_{accum}(V_{ijk}) = \sum_{m=1}^{N} W_m D_m(V_{i'j'k'}) \Big/ \sum_{m=1}^{N} W_m \tag{3}$$

where W_m is the time weight and $D_m(V_{i'j'k'})$ is the dose in the interpolated CT image for breathing level BL_m, and N is the total number of the interpolated CT images used. If equal time weights are used, the accumulated dose distribution will be scaled by the total number of the CT images used. Thus, we have

$$D_{accum}(V_{ijk}) = \sum_{m=1}^{N} D_m(V_{i'j'k'}) \Big/ N \tag{4}$$

3. Results and discussion

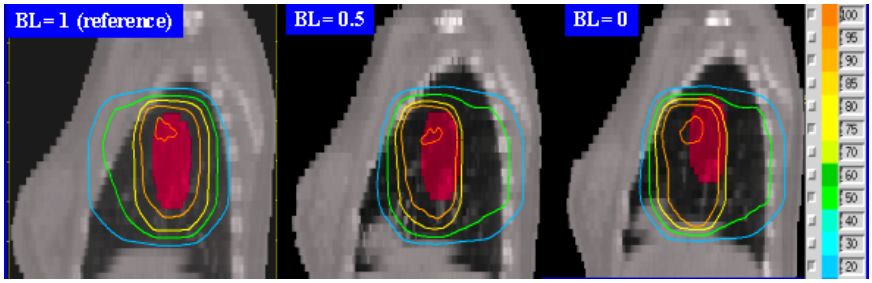
3.1. Breathing motion effect on dose distribution

Figure 2 (a) shows the interpolated CT image of breath level 0.5, obtained from the CT images of the end inspiration (*BL*=1) and the end expiration (*BL*=0). A 0.5cm chest wall movement resulted in a change of ~1cm in the upper portion of the lung, and ~1.5cm for the lower portion of the lung, in the superior-inferior direction. In this example, the tumor target (PTV~22.50cm^3) was assumed in the middle of the lung with a 0.8cm margin to account for lung motion. The central coordinates of the target in the inspiration image were (LAT=5.30cm, AP=3.00cm, SI=-91.13cm), while the corresponding coordinates in the expiration image were (5.33, 3.28, -90.01). The dose distributions were computed for the 4-field 3DCRT plan at different breathing levels *BL*=1, 0.5 and 0. In this demonstration, the end inspiration CT image (*BL*=1) was chosen as the reference CT image, i.e. the CT image for treatment planning and dose accumulation. The five isodose lines represented 100%, 90%, 75%, 50% and 20% of the prescription dose of 55Gy, respectively. As expected, the target (PTV) was entirely included by the 90% isodose line for the reference CT image. However, for *BL*=0.5, part of the target was outside the 90% isodose line, while for *BL*=0, only the 50% isodose line covered the whole target. Figure 2 (b) shows the reference dose distribution (thick lines) and the accumulated distribution (thin lines) on the end inspiration CT image, and the related DVHs for the target and the lung. The accumulated dose was obtained from the dose distributions for breathing levels *BL*=1, 0.8, 0.6, 0.4, 0.2, and 0. According to the sinusoidal distribution, the relative time weights for the corresponding *BL* values were 1.0, 0.28, 0.23, 0.23, 0.28, and 1.0, respectively. We found that the accumulated target dose coverage was worse than the planned one, and there was more lung volume (~10%) receiving a mid-range dose (3-30Gy) in the accumulated dose distribution taking into account the effect of breathing motion.

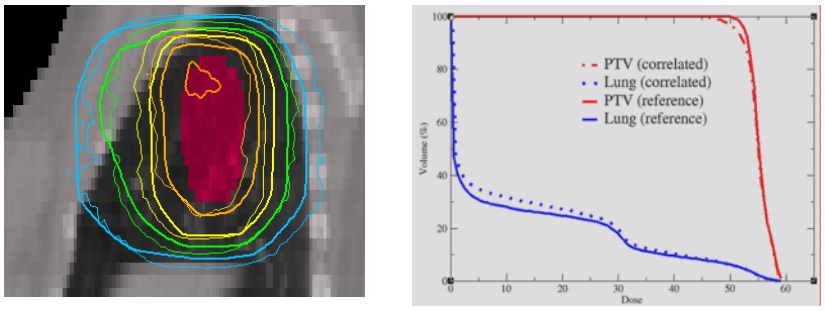
3.2. Breathing motion effect on different planning margins for 3DCRT

Figure 3 shows the breathing motion effect on different planning margins for 3DCRT. Solid lines are the DVHs for the reference CT image at BL=1 (a 1cm margin was used). The breathing cycle was assumed as a sinusoidal curve as shown in Fig.1. Dotted lines are the DVHs for the accumulated dose distribution for a 1cm margin and dashed lines are for a 2cm margin. The dose distributions were accumulated based on patient geometries at BL= 0, 0.25, 0.5, 0.75 and 1, respectively. The results showed that for a 1cm planning margin, the breathing motion would cause cold spots in the target dose coverage. When a 2cm margin was applied, target dose coverage was not be affected by the breathing

motion. However, the lung volume receiving a dose of 5-50 Gy increased by ~ 50% for a tumor in the upper lung, and by up to ~60% for a tumor in the lower lung. Our results indicate that using a larger margin is not the best solution for solving the uncertainties due to breathing motion in lung cancer treatment, especially for patients with tumors in the lower lung. This is consistent with the findings by Allen et al [15]. For this reason, advanced techniques have been developed to compensate for lung motion in radiotherapy [16].



(a)



(b)

Figure 2. Dose accumulation taking into account breathing motion. (a) Dose distributions for patient CT images at different breathing levels, *BL*=1, 0.5 and 0. *BL*=1 was chosen as the reference. (b) Comparison of the reference and accumulated dose distributions and their DVHs. For dose distributions, thick lines are for the reference and thin lines are for the accumulated dose. For DVHs, solid lines are for the reference CT image, dashed and dotted lines are for the accumulated dose distribution.

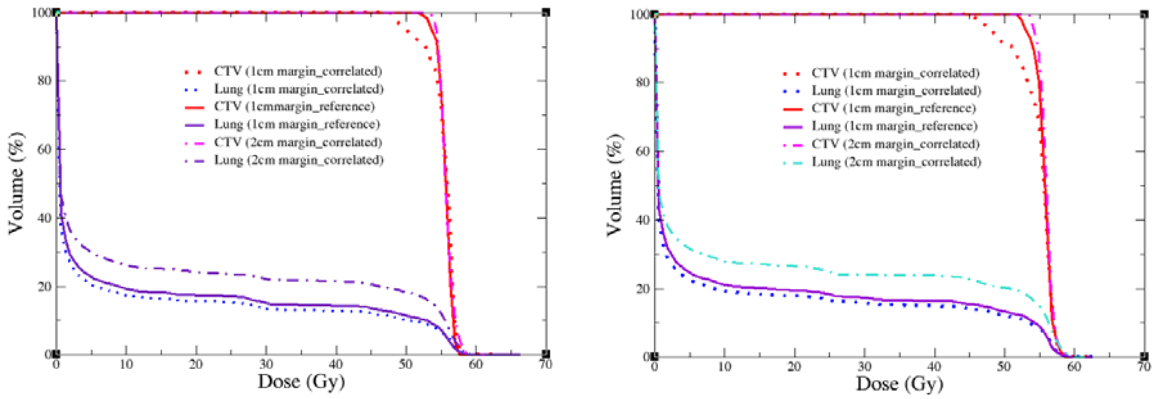


Figure 3. Breathing motion effect for different treatment planning margins: (left) tumor in the lower lung and (right) tumor in the upper lung. Solid lines are the DVHs for the reference CT image (*BL*=1, 1cm margin). Dotted/dashed lines are for accumulated dose distributions with 1cm/2cm planning margins (dose accumulated from *BL*= 0, 0.25, 0.5, 0.75 and 1).

4. Conclusions

We have presented a 4D image registration and dose correlation model for breathing motion studies and demonstrated the use of this 4D planning tool to evaluate the motion effect on the delivered dose for lung cancer treatments with 3DCRT. Our model can create patient simulation geometry at any intermediate breathing phases from CT images taken at the end inspiration and end expiration phases, and establish the voxel correlation between the interpolated CT image and the reference CT image (either the end inspiration or the end expiration CT image) for dose accumulation taking into account the breathing motion effect. Our preliminary results suggested that using a larger margin might not be a good solution for solving the uncertainties due to breathing motion. Our model can be a useful tool for dose calculation and plan evaluation for the implementation of 4D radiotherapy clinically.

References

[1] Keall P, Chen G, Joshi S, Mackie T and Stevens C 2003 Time─ the 4th dimension in radiotherapy *Int. J. Radiat. Oncol. Biol. Phys.* **57** S8-S9
[2] Pan T, Lee TY, Rietzel E and Chen G 2004 4D-CT imaging of a volume influenced by respiratory motion on multi-slice CT *Med. Phys*. **31** 333–40
[3] Brock KK, Balter JM, Dawson LA, Kessler ML, and Meyer CR 2003 Automated generation of a four-dimensional model of the liver using warping and mutual information *Med. Phys*. **30** 1128-1133
[4] Brock KK, Sharpe MB, Dawson LA, Kim SM, and Jaffray DA 2005 Accuracy of finite element model-based multi-organ deformable image registration *Med. Phys.* **32** 1647-59
[5] Rosu M, Chetty IJ, Balter JM, Kessler ML, McShan DL and RKT Haken 2005 Dose reconstruction in deforming lung anatomy: Dose grid size effects and clinical implications *Med. Phys.* **32** 2487-95
[6] Keall P, Siebers J, Joshi S and Mohan R 2004 Monte Carlo as a four-dimensional radiotherapy treatment-planning tool to account for respiratory *Phys. Med. Biol.* **49** 3639-48
[7] Rietzel E, Ghen GTY, Choi NC and Willet CG 2005 Four-dimensional image-based treatment planning: target volume segmentation and dose calculation in the presence of respiratory motion *Int. J. Radiat. Oncol., Biol., Phys*. **61** 1535-50
[8] Schreibmann E, Chen GTY and Xing L 2006 Image interpolation in 4D CT a BSpline deformable registration model *Int. J. Radiat. Oncol., Biol., Phys*. **64** 1537-50
[9] Flampouri S, Jiang SB, Sharp GC, Wolfgang J, Patel AA and Choi NC 2006 Estimation of the delivered patient dose in lung IMRT treatment based on deformable registration of 4D-CT data and Monte Carlo simulations *Phys. Med. Biol.* **51** 2763-79
[10] Heath E and Seuntjens J 2006 A direct voxel tracking method for four-dimensional Monte Carlo dose calculations in deforming anatomy *Med. Phys.* **33** 434-45
[11] Keall P, Joshi S, Vedam S, Siebers J, Kini V and Mohan R 2005 Four-dimensional radiotherapy planning for DMLC-based respiratory motion tracking *Med. Phys.* **32**:942-51
[12] Ding M, Li J, Deng J, Fourkal E and Ma C-M 2003 Dose correlation for thoracic motion in radiation therapy of breast cancer *Med. Phys*. **30** 2520-9
[13] Ma C-M, Li J S, Pawlicki T, Jiang S B and Deng J, Lee M C, Koumrian T, Luxton M and Brain S 2002 A Monte Carlo dose calculation tool for radiation therapy treatment planning *Phys. Med. Biol.* **47** 1671-89
[14] Deng J, Jiang SB, Kapur A, Li JS, Pawlicki T, Ma C-M 2000 Photon beam characterization and modelling for Monte Carlo treatment planning *Phys. Med. Biol*. **45** 411-27
[15] Allen AM, Siracuse KM, Hayman J and Balter JM 2004 Evaluation of the influence of breathing on the movement and modeling of lung tumors *Int. J. Radiat. Oncol. Biol. Phys.* **58** 1251-7
[16] Berbeco RI, Neicu T, Rietzel E et al. 2005 A technique for respiratory-gated radiotherapy treatment verification with an EPID in *cine* mode. *Phys. Med. Biol.* **50** 3669-79