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Image guided RT

Motion compensated digital tomosynthesis

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ABSTRACT

Background and purpose: Digital tomosynthesis (DTS) is a limited angle image reconstruction method for cone beam projections that offers patient surveillance capabilities during VMAT based SBRT delivery. Motion compensation (MC) has the potential to mitigate motion artifacts caused by respiratory motion, such as blur. The purpose of this feasibility study was therefore to develop and evaluate motion-compensated DTS (MC-DTS).

Material and methods: MC-DTS images were reconstructed by back projection of X-ray projection images acquired over 30° arcs. Back projection lines were deformed according to an a priori motion model derived from the 4D planning CT. MC-DTS was evaluated on a respiratory motion phantom and 3 lung cancer patients. Respiratory artifact reduction was assessed visually and quantified by fitting a cumulative Gaussian function to profiles along the background-GTV transition in the CC direction.

Results: MC reconstruction was fast enough to keep up with image acquisition and considerably reduced motion blur visually. Quantitatively, MC reduced the background-GTV transition distance by 49%.

Conclusion: Motion compensation considerably improved the image quality of DTS images of lung cancer patients, giving an opportunity for more accurate DTS guidance and intra-fraction monitoring concurrent with VMAT delivery.

Lung cancer is the most common cause of cancer related death in the western world. For inoperable lung cancer, radiotherapy is the treatment of choice. For inoperable stage I non-small cell lung cancer (NSCLC), Stereotactic Body Radiotherapy (SBRT) has become an excellent treatment option. SBRT is characterized by highly conformal dose distributions with ablative doses delivered in a few fractions [1]. SBRT therefore requires high-precision treatment delivery techniques to limit treatment volumes and thereby toxicity.

Various groups have reported on the use of Cone Beam CT (CBCT) [2] integrated with the linear accelerator to localize the target with high precision [3–5]. Reconstruction of a 3D CBCT requires acquisition over >180°. Since the image modality is mounted to the linac, CB imaging speed during treatment is limited by the gantry speed. Therefore the temporal resolution of surveillance during treatment is limited.

An alternative imaging method for IGRT is digital tomosynthesis (DTS) [6]. DTS is a limited angle image reconstruction method from 2D cone-beam projections which improves the visibility of anatomy compared with radiographic imaging. Due to the limited acquisition angle of DTS, it has the potential to significantly increase the temporal resolution of patient surveillance during VMAT delivery at the cost of reduced resolution in one direction.

Despite the increased imaging frequency of DTS, its acquisition is still slow relative to the typical breathing cycle and therefore DTS images of the thorax exhibit respiratory induced motion artifacts such as blur. Maurer et al. [7] therefore developed four-dimensional DTS to mitigate such motion artifacts but hereby typically induce view aliasing artifacts due to coarse angular sampling.

For various imaging modalities, motion compensated reconstruction has been proposed as an alternative to 4D imaging [8–11] which simultaneously reduces motion artifacts and maintains high contrast to noise ratio's. The purpose of this study was therefore to investigate the feasibility of motion compensated digital tomosynthesis (MC-DTS) aimed at patient monitoring during RT.

Materials and methods

Scanners

Treatment planning was performed on 4D CT scans acquired on a helical CT scanner (Siemens, 24-slice Somatom Sensation Open, Siemens, Forchheim, Germany). A thermocouple placed at the entry of a nasobuccal oxygen mask was used to detect the respiratory signal. The 4D CT scans were reconstructed with 10 frames equally

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distributed in time and a $1 \times 1 \times 3$ mm^3 voxel size (see Wolthaus et al. [10]).

The cone-beam scanner was mounted on a linear accelerator (Synergy 4.6, Elekta Oncology Systems, Crawley, UK). Projection images were acquired at 5.5 fps using pulses of 120 kVp 0.16 mAs. The flat panel detector had a field of view of 26.7×26.7 cm^2 at the plane of the isocentre and images were stored with a pixel pitch of 0.5 mm.

Phantom data

Phantom experiments were performed with the dynamic thorax phantom (CIRS, Norfolk, Virginia, USA). The phantom approximates an average human thorax shape, with lung and vertebrae shaped inserts made of lung and bone tissue equivalent material, respectively. A water equivalent 3 cm diameter sphere representing a tumor was inserted off-axis into a lung tissue equivalent cylinder which was embedded in the lung tissue representing part of the phantom. Respiration like motion of the target was programed according to the Lujan model [12] trough translations and rotations of the cylinder.

In CC direction a Lujan model with $n = 3$ (sin^6) and a peak-to-peak amplitude of 2 cm was applied. In AP direction, a sinusoidal motion pattern ($n = 1$) with a peak-to-peak amplitude of 1 cm was applied. Projection images were acquired over a 360° arc with a gantry rotation speed of 90°/min resulting in a total number of 1317 projections. The isocentre was placed approximately in the center of mass of the spherical insert.

Patient data

Three non-small-cell-lung-cancer (NSCLC) patients with peripherally located tumors (<5 cm diameter), treated with SBRT (3×18 Gy) were included in this study. We selected patients from our SBRT database with different tumor sizes, who were recently treated and showed considerable tumor motion on pre-treatment 4D CBCT (see Table 1). The patients received an online 4D CBCT scan prior to treatment to align the target with the planned position. The CBCT was acquired over 200° with a gantry rotation speed of 75°/min yielding approximately 900 projection images.

DTS images were reconstructed for both phantom and patient scans with and without motion compensation.

DTS reconstruction

DTS images were reconstructed from projection series over an arc $\geqslant$200° with an in-house implementation of the FDK algorithm [13,14]. In anticipation of efficient online analysis, DTS images were first reconstructed over 10° arcs and stored in a 4D matrix, with the 3 spatial axes, x, y, z and gantry angle as a fourth dimension. To obtain a 30° DTS image, 3 consecutive 10° images were averaged. All DTS images were reconstructed at a $256 \times 256 \times 256$ grid with a $1 \times 1 \times 1$ mm^3 voxel size. Note that the average over all DTS images represents a normal CBCT reconstruction, with the exception of the first and last DTS images that require parker weighting [15].

Table 1
Peak-to-peak amplitudes of the respiratory induced target motion for the patients and the phantom as well as the target volumes.

	Peak-to-peak amplitude on 4D CBCT (cm)				GTV (cm^3)
	LR	CC	AP	Vector length	
Phantom	0.1	2.0	1.0	2.24	37.7
Patient 1	0.03	0.79	0.25	0.83	27.7
Patient 2	0.39	1.10	0.53	1.28	4.6
Patient 3	0.21	1.48	0.19	1.51	1.8

A priori motion model

A model of the patient respiratory motion was estimated a priori from the 4D planning CT and subsequently used in motion compensated DTS reconstruction. Motion modeling and motion compensated CBCT reconstruction have previously been described in detail by Wolthaus et al. [16] and Rit et al. [11], and are summarized in the sections below.

A phase based optical flow algorithm was used to register all phases of the 4D scan to the exhale phase producing a 4D deformation vector field (DVF). Subsequently, the average DVF was subtracted from each phase (including the zero DVF for the exhale phase) to obtain a model relative to the time weighted mean position, i.e., the Mid-Position.

The deformable registration algorithm did not produce a reliable motion model for the phantom data, due to its piece-wise constant density construction. Therefore, a motion model for the phantom scan was manually constructed based on the programed 3D target motion. For simplicity, the DVF was set constant for the whole image (i.e., a rigid motion model was used for the phantom), such that steady parts of the phantom show motion blur in the motion-compensated reconstruction.

Motion compensated DTS reconstruction

Motion-compensated DTS reconstruction is performed by back projection over curved lines, i.e., the acquisition lines corresponding to X-rays deformed to the mean position. To that end, first the projection images are sorted according to their breathing phase using the Amsterdam shroud technique [17]. For each projection, after determination of the breathing phase, the motion model for that phase was applied during back projection into the corresponding 10° DTS cube. However, since the phase signal from the planning CT is measured with an external monitoring system, and an internal respiratory motion signal is used for the DTS reconstruction, there is a need for calibration. This was achieved by taking the average displacement over the 4D planning CT derived DVF per phase in cranial–caudal direction, effectively measuring the internal signal of the 4D CT. Subsequently, the phase of this average signal is determined and used to map the Amsterdam shroud signal to the phase of the DVF. Note that by deforming the back projection geometry, there is a risk of effectively displacing the anatomy. Therefore, the phase is defined as linear in time per respiratory cycle. Consequently, every phase is applied as often, which in combination with a zero mean displacement of the 4D-DVF derived from the planning CT, as described above guarantees that the average displacement over the breathing cycles is zero.

To optimize computational efficiency of the reconstruction, curved back projection lines from the motion model were approximated by piece wise linear subsections with a maximal deviation of half a voxel dimension. These subsections were generated once prior to the first image acquisition for that patient.

Image quality analysis

To quantify the deblurring effect of MC-DTS reconstruction, a profile comparison method was applied. To that end, the gray-values of a 30° DTS slice in cone-beam's eye view were extracted for both the DTS and MC-DTS images. Average gray-values were extracted along the CC direction over a 6 mm wide band through the center of the GTV/spherical insert. Subsequently, the following modified cumulative normal distribution (Z-function) was fitted to the gray value profile:

$$Z = A + B * \frac{1}{2} erfc\left(\frac{\mu - x}{\sqrt{2}\sigma}\right) \tag{1}$$