Scientific Article

Model evaluation of rapid 4-dimensional lung tomosynthesis

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Abstract

Purpose: This is an investigation of a lung motion digital tomosynthesis (DTS) model using combined stationary detector and stationary cold cathode x-ray sources at projection acquisition rates that exceed the present norms. The intent is to reduce anatomical uncertainties from artifacts inherent in thoracic 4-dimensional computed tomography (CT).

Methods and materials: Parameters necessary to perform rapid lung 4-dimensional DTS were studied using a conventional radiographic system with linear motion of the x-ray source and a simple hypothetical hardware performance model. Hypothetical rapid imaging parameters of sweep duration, projections per second, pulse duration, and tube current (mA) were derived on the basis of 0.5 mm and 1 mm motion captures per phase, 10 and 15 breaths per minute (bpm), 10 to 40 mm breathing amplitude, and 2 signal-to-noise ratio (SNR) levels. Anterior-posterior and lateral projection images of a normal size anthropomorphic thorax phantom with iodine contrast inserts were collected and reconstructed with an algebraic algorithm to study the effects of reduced x-ray output associated with field emission cold cathodes composed of carbon nanotubes or metal Spindt-type. Radiographic projections were collected at 3 SNR levels that were set at standard clinical DTS milliampere-seconds (mAs) and reduced corresponding to 50% and 25% standard DTS mAs to simulate a reduced x-ray output.

Results: The DTS SNR of the inserts was superior in all reconstructions at clinical mAs versus automatic exposure-control radiographs and superior in 3 of 4 at the 50% and 25% mAs levels. The most demanding performance parameters corresponding to 40 mm amplitude, 15 bpm, 0.5 mm motion capture limit, and 61 projections were sweep duration (10.4 msec), projection rate (5862 projections per second), pulse duration (0.161 msec), current 189 mA anterior-posterior, and 653 mA lateral.

Conclusions: Feasibility depends on the output performance of stationary cold cathode hardware being developed for DTS. Present image receptor technology can accommodate frame acquisition rates.

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Introduction

Stereotactic body radiation therapy (SBRT) requires accurate delineation of the internal target volume. Lung SBRT simulation involves 4-dimensional computed tomography...
Four-dimensional CT projections per Four-dimensional CT will continue to provide. Maltz et al published a fixed-phase digital tomosynthesis (DTS) reconstruction technique that merges phase-resolved maps of initial 4-dimensional CT scans with carbon nanotube cathodes.

Hsieh and NG proposed linac gantry-mounted DTS to provide real-time guidance at up to 30 frames per second. The application of DTS to nodule detection, phase-matched, digitally reconstructed radiographs and artifact suppression relative to 4-dimensional CT. Zhang et al proposed lung localization during linac radiation therapy via DTS reconstructions from 4-dimensional CT simulation, phase-matched, digitally reconstructed radiographs and DTS images from cone beam CT collected at treatment. Their study captured 10 breaths per minute (bpm), end of exhalation images at 30 projections per second (pps), 80 kVp, 5 mA anode current, 25 and 50 milliseconds per projection pulse, and 5 projections per respiration cycle with the whole sequence requiring 36 seconds across 6 respiration cycles.

Studies have been published on radiation therapy guidance DTS. Maurer et al proposed linac on-board 4-dimensional DTS using respiratory-correlated cone beam CT projections and found greater craniocaudal motion artifact suppression relative to 4-dimensional CT. Hsieh and NG proposed linac gantry-mounted DTS to provide real-time guidance at up to 30 frames per second. Shan et al focused on prospective gated chest DTS using stationary carbon nanotube x-ray sources.

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Lung digital tomosynthesis (DTS) has been investigated as a modality for pulmonary nodule detection and management. However, nodules are not necessarily indicative of cancer and play no role in SBRT target delineation. The application of DTS to nodule detection lends support to inclusion in treatment planning. Investigations into stationary x-ray sources for gated chest DTS were conducted at the University of North Carolina at Chapel Hill. Shan et al focused on prospective gated chest DTS using stationary carbon nanotube x-ray sources. Their study captured 10 breaths per minute (bpm), end of exhalation images at 30 projections per second (pps), 80 kVp, 5 mA anode current, 25 and 50 milliseconds per projection pulse, and 5 projections per respiration cycle with the whole sequence requiring 36 seconds across 6 respiration cycles.

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In 4-dimensional CT, amplitude and rate variations coupled with scan volume limitations leads to artifacts. For example, inhalation rate or amplitude may vary between different sections of the lung in the craniocaudal direction as the z-limited beam passes over. Lung hysteresis is affected by differences in breathing amplitudes and rates, all of which are patient specific and variable, making tumor motion prediction difficult. The working premise in rapid 4-dimensional DTS is that spatiotemporal uncertainties in 4-dimensional CT can be reduced or eliminated by single-cycle imaging of the entire lung, provided the patient can perform a single good breathing cycle.

Rapid 4-dimensional DTS will limit motion artifacts and provide high resolution in the coronal and sagittal reconstruction planes. Spatial resolution in the reconstruction plane will depend on geometric blur and flat-panel detector size. In-plane resolution in the coronal and sagittal planes can facilitate precise target segmentation. The approximate 1 cm slice thickness inherent in DTS can be compensated by orthogonal projections, with Boolean merger of segmentations from orthogonal image sets compensating for lower resolution in the orthogonal depth direction. Four-dimensional CT will continue to provide curvilinear contour accuracy in the axial plane, complemented by rapid 4-dimensional DTS. Target conspicuity may be enhanced through intravenous iodine contrast as administered for 4-dimensional CT.

Operation of this model would entail 61 projections through a linear sweep angle ≥40°. The sources configuration and sweep orientation can be 2-dimensional and the number of projections negotiable. Sweep refers to the sequential firing of all focal spots during the collection of a complete set of tomographic projections at a given single breathing phase. A larger number of projections minimizes the ripple artifact, and a greater sweep angle improves depth resolution. Tube current milliamperes-seconds (mAs) per projection over a single arc are a fraction of the mAs required for a conventional radiograph. Projections per second per phase is set to limit the captured motion during any phase to a predefined limit. Heart motion artifacts may be reduced by simultaneous cardiac gating with imaging that is performed during the quiescence, although this would extend imaging beyond a single breathing cycle.
system sharing by physical link between the CT and 4-dimensional DTS system will allow coregistration of 4-dimensional DTS and 4-dimensional CT, with lesion selection in the CT image set determining the target of 4-dimensional DTS reconstruction and lesion delineation. Four-dimensional CT will continue as the primary modality and provide axial curvilinear contouring accuracy.

Stationary sources rapid exposures are provided by cold cathode carbon nanotube field emission or higher output Spindt-type metal field emission cathodes.

High frame rates will not challenge the x-ray emitter repetition rate because each will fire once per sweep, with 10 or fewer sweeps over a breathing cycle. The frame rate will challenge image receptor and readout systems but is within the limits of available imaging systems (Radiation Monitoring Devices, Inc., Watertown, MA).

Methods and materials

Four-dimensional tomosynthesis model

Tomosynthesis projection pps adequate for limiting craniocaudal motion to a predetermined distance during capture of an individual phase depends on the following:

1) Motion amplitude \( b \)
2) The desired maximum tissue travel during the sweep \( D \);
3) The number of projections per sweep \( NP \); and
4) Phase-dependent tissue velocity.

Image quality depends on the number of projections, sweep angle, and mAs per projection.

Lujan et al describe the following breathing motion model:

\[
z(t) = z_0 - b \times \cos^{2n}(\frac{\pi t}{\tau} - \phi),
\]

where \( z(t) \) is craniocaudal position at time \( t \); \( z_0 \) is the position at exhale; \( b \) is the amplitude of motion; \( z_0 - b \) is the position at inhale; \( \tau \) is the period of the breathing cycle; \( n \) is the parameter determining the waveform steepness and flatness; and \( \phi \) is the starting phase of the breathing cycle.

The instantaneous velocity at time \( t \) is:

\[
\frac{dz(t)}{dt} = \frac{2n \pi b}{\tau} \cos^{2n-1}(\frac{\pi t}{\tau}) \sin(\frac{\pi t}{\tau}).
\]

The sweep duration per phase depends on the velocity at which the collection is performed. The required duration of a sweep at time \( t_A \) can be computed as:

\[
Sweep \, Duration(t_A) = D \left[ \frac{dz(t)}{dt} \right]_{t_A}.
\]

where \( t_A \) is the midpoint time of image acquisition, \( t_A = \frac{\tau}{\pi} (\phi_A + \phi_b) \), and \( \phi_A \) is the breathing phase at acquisition. Time duration of acquisition extends from

\[
t_A - D \left[ \frac{dz(t)}{dt} \right]_{t_A} \leq t_A + D \left[ \frac{dz(t)}{dt} \right]_{t_A}.
\]

Craniocaudal lung tissue velocity is greatest, and sweep duration least, midway between maximum inhalation and maximum exhalation, in which:

\[
\cos^{2n}(\frac{\pi t}{\tau} - \phi) = 0.5.
\]

The required pps at time \( t \) can be computed as:

\[
Frame \, Rate = \frac{NP}{Sweep \, Duration}.
\]

mAs per projection is

\[
mAs \, per \, Projection = \frac{(mAs_{AEC}) \times f}{NP}.
\]

where \( mAs_{AEC} \) is the mAs per single radiographic exposure as determined by the automatic exposure control, and \( f \) is a noise-scaling factor with 10 regarded as clinically acceptable.

\[
Pulse \, Duration = \left( \frac{1}{Frame \, Rate} \right) - frame \, gap,
\]

where frame gap is the time duration between consecutive projections,

\[
Frame \, Gap = \left( \frac{1}{100,000 \, fps} \right) = 0.01 \, msec,
\]

the 100,000 pps basis for the frame gap is short of the maximum available rate for radiographic imaging of 150,000 pps (Radiation Monitoring Devices, Inc.) and

\[
mA = \frac{mAs \, per \, projection}{Pulse \, Duration}.
\]

Imaging system

The GE Brivo XR385 digital radiographic imaging system with stationary Gd2O2S:Tb receptor, 16 bits per pixel, 2022 × 2022 matrix, 38.42 × 38.42 cm² (GE Healthcare, Aurora, OH) was used. The grid ratio is 13:1 and 70 lines/cm. The source-detector distance is 110 cm at 0° projection.

Phantom

The Kyoto Kagaku Chest Phantom N1 (Kyoto Kagaku, Torrance, CA) was used in this study. Iodine-infused lesions
were simulated with microcentrifuge tubes of 1.5 mL volume, 10 mm inner diameter, and a ratio of 30:1 solution of water to Omnipaque (GE Healthcare, Aurora, OH) contrast media for a predilution concentration of 350 mg/mL. The ratio was based on a prescribed injection of 70 mL at 3 mL/sec and cardiovascular output of 5000 mL/min:

$$\frac{\left(\frac{5000 \text{ mL}}{\text{min}}\right) \times \left(\frac{1 \text{ min}}{60 \text{ sec}}\right) \times \left(\frac{70 \text{ mL}}{3 \text{ mL/sec}}\right)}{70 \text{ mL}} = 27.8 = 30 \quad (11)$$

The tubes were positioned in the right and left lung posterior to the heart.

**Projection imaging**

To simulate a cold cathode emitter, focal spot linear array, the x-ray tube was translated along the craniocaudal direction. The tube angle was adjusted at each projection so that the central ray extended from the focal spot to the center of the image receptor. Tube positions and angles were verified using a Digi-Pas model DWL-80G digital angle gauge (Digipas USA LLC, Avon, CT) and Kobalt LDM-35 laser distance meter (Lowe’s Corp., Mooresville, NC). Radiographic automatic exposure-control (AEC) settings were average size chest AP and lateral, 120 kVp, zero added filtration.

**Signal-to-noise ratio and milliampere-seconds relation**

Projection mAs was computed per equation (3) using $f = 10, 5, \text{and} 2.5$. The minimum mAs setting for the generator exceeded computed values; therefore, the projection image pixel intensity was reduced and noise added to simulate computed mAs values using image processing software Fiji in the following order:

$$(N_f)_{i,j} = \left(\frac{\text{mAs per projection}}{\text{mAs}_{\text{raw}}}\right) \times (N_{\text{raw}})_{i,j}, \quad (12)$$

followed by

$$\sigma_f = \sigma_{\text{modified}} \times \sqrt{\frac{\text{mAs}_{\text{raw}}}{\text{mAs per projection}}}, \quad (13)$$

where $(N_f)_{i,j}$ are the pixel values in the projection image modified per $f$; $(N_{\text{raw}})_{i,j}$ are the pixel values in the original raw projection image; mAs per projection is per eq (3); mAsraw is the manual mAs setting used to produce the raw projection image; $\sigma_f$ is the pixel standard deviation of a region of interest (ROI) over a physically uniform section of the phantom that is inferior to the diaphragm in the final image; $\sigma_{\text{modified}}$ is the pixel standard deviation of an ROI over a uniform section of the phantom beneath the diaphragm in the raw projection image modified per equation (12).

**Image processing and reconstruction**

Tomosynthesis reconstruction was done with a C++ algebraic algorithm. Noise injections and image analyses were done with Fiji.

**Image analysis**

Iodine SNR levels were determined as:

$$\text{SNR} = \frac{(\text{ROI mean})_I - (\text{ROI mean})_{\text{Bkg}}}{(\text{ROI std.dev.})_{\text{Bkg}}} \quad (14)$$

where $(\text{ROI mean})_I$, $(\text{ROI mean})_{\text{Bkg}}$, and $(\text{ROI std.dev.})_{\text{Bkg}}$ are the ROI mean values for the iodine insert and background regions and the standard deviation in ROI values in background regions, respectively. ROI locations were consistent from image to image.

**Results**

**Technique factors**

AEC radiographic, small focal-spot techniques were 120 kVp, AP projection 0.37 mAs, and lateral projection 1.28 mAs. Manual techniques per projection were 120 kVp, zero added filtration for AP and lateral projections, and 0.43 mAs, which is referred to as full mAs.

Tomosynthesis model results per equations (1) through (10) using $n = 3$ are given in Table 1. Anode currents versus breathing amplitude $b$ at $f = 5, \text{bpm} = 10 \text{ and} 15$, and $D = 0.5$ and 1 mm are presented in Figure 1 for AP and lateral views. Dependence relationships in the model are given in Table 2. Figure 2 presents scout views and DTS reconstructions.

**Iodine signal-to-noise ratio comparisons**

SNR results are presented in Table 3. SNR values vary in an anthropomorphic lung phantom due to anatomical noise. Generally, tomosynthesis improves SNR over a single radiograph when anatomical noise is present.

**Conclusions and discussion**

Imaging parameter relations are provided in Tables 1 and 2 and Figure 1. Proportional relations exist between mA and both $D$ and bpm. Lateral imaging requires increased anode current.
Current and pulse duration demands present engineering challenges. Current density (A/cm²) of cold cathodes varies inversely with emitter surface area. Smith et al reported a carbon nanotube field-emission minimum pulse duration of 0.02 msec and maximum current density of 0.8 A/cm² from a 0.04 cm² surface area. Schwoebel et al achieved a 300 mA current with a 0.1 msec pulse width using a 50,000 tip array Spindt-field emitter array with expectations of >1 A from the same size array. Demands might be met by focusing beams from ≥2 emitters onto each focal spot. Relative SNR was chosen to test mAs reduction. DTS SNR is also heavily dependent on anatomical noise. Projection mAs was reduced from clinical standard values using \( f = 2.5 \) and 5. Figures 2 and 3 and Table 3 support DTS in 3 of 4 capsule views using \( f = 5 \) and demonstrate that objects with SNR = 2 are visible at least in this configuration. Lateral scout AEC radiographic images show lung capsule visibility as absent but present in the DTS reconstructions. AP behind heart DTS SNR is less than the scout AEC projection SNR at \( f = 5 \) and 2.5 due to the simulated noise increase at reduced mAs, which overwhelmed signal improvement provided by DTS. Generally, slice sensitivity profiles produced with DTS exhibit a nonuniform frequency response with an inverse relationship between slice profile thickness and spatial frequency. The abundance of low-frequency information in the posterior portion of the heart nearest to the capsule contributed to the capsule plane. Nonuniform frequency response can be improved with refined reconstruction techniques.

Ripple artifacts and poor visibility of the lateral right-lung capsule reconstructions result from a significant amount of information falling outside the detector due to projection geometry. This can be corrected by increasing the detector length in the sweep direction or distributing sources in a 2-dimensional pattern and widening the detector to ensure that an adequate number of projections is included in the reconstruction and by positioning the lateral image receptor on a unilateral lesion side.

### Table 1: Sample technique factors required by 4-dimensional Tomo model

| D (mm) | b (mm) | Projection rate (sec⁻¹) | Sweep duration (msec) | Pulse duration (msec) |
|--------|--------|-------------------------|-----------------------|-----------------------|
| 0.5    | 15     | 40                      | 5862                  | 10.41                 | 0.161                 |
|        |        |                         | 4397                  | 13.87                 | 0.217                 |
|        |        |                         | 2931                  | 20.81                 | 0.331                 |
|        |        |                         | 1466                  | 41.62                 | 0.672                 |
|        | 10     | 40                      | 3908                  | 15.61                 | 0.246                 |
|        |        |                         | 2931                  | 20.81                 | 0.331                 |
|        |        |                         | 1954                  | 31.22                 | 0.502                 |
|        |        |                         | 977                   | 62.44                 | 1.014                 |
| 1      | 15     | 40                      | 2931                  | 20.81                 | 0.331                 |
|        |        |                         | 2198                  | 27.75                 | 0.445                 |
|        |        |                         | 1466                  | 41.62                 | 0.672                 |
|        |        |                         | 733                   | 83.25                 | 1.355                 |
|        | 10     | 40                      | 1954                  | 31.22                 | 0.502                 |
|        |        |                         | 1466                  | 41.62                 | 0.672                 |
|        |        |                         | 977                   | 62.44                 | 1.014                 |
|        |        |                         | 489                   | 124.87                | 2.037                 |

b, motion amplitude; bpm, breaths per minute; D, desired maximum tissue travel during the sweep.

### Table 2: Dependence relations among 4-dimensional Tomo model parameters

| D     | b     | NP   | f   |
|-------|-------|------|-----|
| mAs/projection | P     | IND  | NP  | f   |
| Projection rate  | IP    | P    | P   | IND |
| Sweep duration   | P     | IP   | IND | IND |
| Pulse duration   | P     | IP   | IND | IND |
| mA               | IP    | P    | IND | P   |

b, motion amplitude; D, desired maximum tissue travel during the sweep; f, noise-scaling factor with 10 regarded as clinically acceptable; IND, independent; IP, inversely proportional; mAs, milliampere-seconds; NP, number of projections per sweep; P, proportional.
Patient-specific anatomical noise limits quality metric predictability. An appropriate approach might be mAs adjustment on the basis of relationships between a standardized phantom metric and receptor dose during scout exposures. For example, Svalkvist and Bath successfully developed a model to simulate system noise in reduced dose DTS projections on the basis of noise power spectrum and detective quantum efficiency using flat-field images at multiple projections and dose levels. This could be modified to establish patient-specific mAs values as a function of image receptor dose determined during the scout. Exposure versus SNR relations might be established using a standardized phantom of variable size with appropriate test inserts.

Similarly, Li and Dobbins developed a noise power spectrum-based method to simulate reduced exposure and

**Table 3** SNR results at 3 tomosynthesis imaging mAs levels

|                  | Lung capsule SNR | Capsule behind heart SNR |
|------------------|------------------|--------------------------|
| **Anterior-posterior** |                  |                          |
| Tomosynthesis f = 2.5 | 3.86 ± 0.88      | 1.20 ± 0.44              |
| Tomosynthesis f = 5    | 7.60 ± 0.90      | 2.60 ± 0.58              |
| Tomosynthesis f = 10   | 14.11 ± 0.96     | 5.84 ± 0.66              |
| Automatic exposure control | 3.05 ± 0.12      | 3.60 ± 0.15              |
| **Lateral**          |                  |                          |
| Tomosynthesis f = 2.5 | 0.73 ± 0.23      | 2.73 ± 0.26              |
| Tomosynthesis f = 5    | 1.12 ± 0.22      | 5.65 ± 0.87              |
| Tomosynthesis f = 10   | 1.99 ± 0.21      | 8.08 ± 0.74              |
| Automatic exposure control | Not Visible      | 0.83 ± 0.03              |

mAs, milliampere-seconds; SNR, signal-to-noise ratio.

**Figure 2** Row A: Anterior-posterior and lateral scout views; row B: f = 2.5 reconstructions; row C: f = 5 reconstructions; row D: f = 10 reconstructions. Columns left to right are capsule behind heart anterior-posterior, capsule behind the heart lateral, capsule in the right lung anterior-posterior, and capsule in the right lung lateral. Red arrows point to capsules in each orientation.
measure SNR versus cumulative exposure for simulated nodules and simulated noise in clinical DTS projections but without actual reduced-dose imaging.²⁸

The \( f \) versus measured SNR, \( \text{SNR}/\text{SNR}_{\text{scout}} \) example could be a format to present data that are generated using the method by Svalkvist and Bath applied to a standardized phantom with inserted test objects.

Additionally, mAs per projection and mA depend on geometry, habitus, x-ray spectrum, subject contrast, and detector efficiency. mA demands may be reduced by improved detectors, reconstruction algorithms that further reduce overlapping structures interference, photon counting (Radiation Monitoring Devices, Inc.), and dual-energy imaging, albeit with increased demands on mA, pulse duration, image detection, and readout.²⁶,²⁹-³⁴

Limitations of depth resolution can be corrected by orthogonal pair imaging coupled with a Boolean combination of contours from both planes, in which case motion capture constraints are best met by simultaneous AP/lateral imaging. This might be facilitated by employing the radiographic and cone beam CT scatter correction method by Siewerdsen et al.³⁵

Using the GE Definium 8000 VolumeRAD DTS system, Sabol determined an average adult male chest examination effective dose of 0.124 to 0.134 mSv, which is less than 75% of that predicted by linear scaling of the mAs ratio, using a 120 kVp spectrum, no additional filtration, and GE-recommended mAs, or noise, scaling factor of 10.³⁹

The reduction was due to focal-spot-to-skin-distance variations, collimation changes with projection angle, rounding down the mAs, and variations in organ exposure for each view. This approximates the effective dose of a single rapid 4-dimensional DTS phase capture. Multiple phases or orthogonal view tomosynthesis would increase the effective dose. Multiphase studies would incur a dose that is approximately equal to:

\[
\text{Effective Dose} \approx 0.75 \times \left( \frac{\text{number of phases}}{\text{mAs scaling factor}} \right) \times \text{(single radiograph dose)}
\]

Ten phases that use an mAs scaling factor of 10 produces an approximate effective dose \( \leq 75 \) single projection radiographs.

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