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A MONTE CARLO BASED MODEL FOR SIMULATION OF DIGITAL CHEST TOMOSYNTHESIS

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The aim of this work was to calculate synthetic digital chest tomosynthesis projections using a computer simulation model based on the Monte Carlo method. An anthropomorphic chest phantom was scanned in a CT scanner, segmented and included in the computer model to allow for simulation of realistic high-resolution x-ray images. The input parameters to the model were adapted to correspond to the VolumeRAD chest tomosynthesis system from GE Healthcare. Sixty tomosynthesis projections were calculated with projection angles ranging from +15 to -15 degrees. The images from primary photons were calculated using an analytical model of the anti-scatter grid and a pre-calculated detector response function. The contributions from scattered photons were calculated using an in-house Monte Carlo-based model employing a number of variance reduction techniques such as the collision density estimator.

Tomographic section images were reconstructed by transferring the simulated projections into the VolumeRAD system. The reconstruction was performed for three types of images using: (1) noise free primary projections, (2) primary projections including contributions from scattered photons and (3) projections as in (2) with added correlated noise. The simulated section images were compared to corresponding section images from projections taken with the real, anthropomorphic phantom from which the digital voxel phantom was originally created.

We here present a work in progress aiming towards developing a model intended for optimization of chest tomosynthesis, allowing for simulation of both existing and future chest tomosynthesis systems.

INTRODUCTION

In planar chest radiography, the projected anatomy is the single most important limiting factor for the detection of subtle nodules^{1,2}. Digital chest tomosynthesis has the potential to increase the detectability of subtle nodules, since it renders tomographic section images and therefore makes it possible to see behind obscuring anatomical structures³. In addition, such examinations can be performed at an effective dose comparable to conventional planar chest radiography, hence significantly less compared to a corresponding examination with chest CT⁴. Nevertheless, because of the relative novelty of the technique, much optimization work is still required in order to support a broader clinical utilization of the technique.

Pineda *et al*⁵ used a computational model to create chest radiographs and chest tomosynthesis images from CT-images. Human as well as model observers were used to evaluate section images produced with different tomosynthesis angles. The section images were compared to planar radiography images. Their conclusion was that very small tomosynthesis angles

were sufficient to outperform planar radiography. Godfrey *et al*⁶ studied optimization of matrix inversion tomosynthesis. They used an anthropomorphic chest phantom and searched for the optimal combination of scan angle, number of input projections and number of tomographic section images. The results from their study suggest a total scan angle of 20 degrees, 71 input projections and 69 reconstructed section images as the optimal parameter combination. The works by these groups are early and important contributions to the optimization of digital chest tomosynthesis. Nevertheless, much work remains to be done in the optimization of chest tomosynthesis. For example, the model by Pineda *et al* does not take the effect of scattered radiation and quantum noise into account. In the study by Godfrey *et al*, the images were collected experimentally, which gives less flexibility compared to producing the images with a computer model.

The present paper describes a work in progress aiming towards developing a model based on the Monte Carlo method. It has potential to provide a highly flexible model for optimization of digital chest tomosynthesis.

MATERIALS AND METHODS

Anthropomorphic voxel phantom

An anthropomorphic chest phantom (Kyoto Kagaku PBU-X-21) was scanned in a Siemens Sensation 64 CT scanner at the tube voltage 120 kV using the reconstruction kernel B31s. A voxel phantom was created using threshold segmentation with a step density function⁷ and included in the computer model, allowing for simulation of realistic high-resolution primary projections. In order to speed up the Monte Carlo simulations, a coarser version of the voxel phantom was created for the calculation of scatter projections. The high-resolution voxel phantom was an array of dimension 232x512x728 voxels ($0.97 \times 0.97 \times 0.6 \text{ mm}^3$) while the low-resolution phantom had the dimensions 58x128x182 ($3.9 \times 3.9 \times 2.4 \text{ mm}^3$). Each voxel was assigned to an index, corresponding to a material composition and mass density that matches the CT number.

Simulation geometry

The simulation geometry is displayed in figure 1 and the main simulation parameters are shown in table 1. The input parameters in the model were adapted to correspond to the VolumeRAD chest tomosynthesis system (GE Healthcare, Chalfont St. Giles, UK). Sixty tomosynthesis projections were calculated with angles ranging from +15 to -15 degrees. In addition, corresponding real phantom tomosynthesis projections were acquired on the VolumeRAD system.

Table 1. Parameters used in the simulations

Parameter	
Tube voltage	120 kV
Total filtration	3 mm Al + 0.1 mm Cu
Focus detector distance	180 cm
Grid ratio	13
Grid strip frequency	70 cm^{-1}
Inter space material	Al
Detector material	CsI
Detector thickness	Undisclosed, estimated as $550 \mu\text{m}$

Primary projections

The images due to primary photons were calculated analytically with the resolution 2022×2022 pixels (pixel size 0.2 mm) using an analytical model of the anti-scatter grid and a pre-calculated detector response function⁸.

Scatter projections

The Monte Carlo simulation model used for the calculation of scatter projections is based on a previous in-house model for planar radiography described in

Sandborg *et al*⁹ and Ullman⁸. The Monte Carlo model simulates photon transport from a point source and through an anthropomorphic voxel phantom, anti-scatter grid and into the image detector. The main improvement, in order to generalize the model to tomosynthesis, was to modify the solid angle for incoming photons, to allow for different angles between the point source and the detector plane. The contributions from scattered photons for each tomosynthesis projection were calculated in a grid of 40×40 points employing a number of variance techniques such as Russian roulette¹⁰ and the collision density estimator¹¹. For each grid point, four different quantities were calculated: (1) an estimate of the mean energy imparted per unit surface area of the detector from scattered photons, ϵ_s , (2) an estimate of the scatter-to-primary ratio, ϵ_s/ϵ_p , (3) an estimate of the signal-to-noise ratio per pixel, SNR_p , and (4) an estimate of air collision kerma. The relative uncertainty in the calculations was 2% (1 S.D.).

Image post processing

The quantities calculated in the Monte Carlo simulations were interpolated from 40×40 points to 2022×2022 points using a bilinear interpolation routine in MATLAB. The quantities were normalized to correspond to the kerma area product (P_{KA}) in the real phantom projections. The P_{KA} was approximately 0.01 Gy cm^2 per projection, i.e. approximately 0.6 Gy cm^2 in total. The scatter projections, ϵ_s , were added to the primary projections and the scatter-to-primary ratios, ϵ_s/ϵ_p , were used to calculate normalization factors for the primary projections. The noise power spectrum, NPS, was measured for the VolumeRAD system. Quantum noise was added to the simulated projections with spatial correlation according to the measured NPS and amplitude according to the calculated SNR_p . The noise addition method is described elsewhere^{8,12}.

Reconstruction

Tomographic section images were reconstructed by transferring the simulated projections into the VolumeRAD system. The reconstruction was performed for three types of images using: (1) noise free primary projections, (2) primary projections including contributions from scattered photons and (3) projections as in (2) with added correlated noise. The simulated section images were compared to corresponding images taken with the real, anthropomorphic phantom from which the digital voxel phantom was created. The nominal slice thickness was 5 mm .

RESULTS AND DISCUSSION

Figure 2 shows the real phantom (2a) and simulated (2b) tomosynthesis projections at the tomosynthetic angle 15 degrees. The positions of the most important anatomical structures such as heart, spine and lungs make an approximate match in the simulated and real images. The spine is more visible in the simulated image. This could be due to inaccurate bone cross-sections or that the ϵ_s/ϵ_p is underestimated in lower mediastinum. Since the segmented phantom has the voxel size $0.97 \times 0.97 \times 0.6 \text{ mm}^3$, the simulated images lack the high frequency structures, such as the smallest vessels that are present in the real phantom image. Small vertical artifacts are visible in the simulated projections. The background of these artifacts is discussed below. Let the y-direction be the direction from the x-ray focus to the center of the image detector. Since the radiation field is relatively parallel to the y-components of the voxels in the central parts of the image, the primary photons are traced through voxels that are projected to the same or close positions in the image. Since the projected voxels have a rectangular shape, this will create vertical structures that are more visible in the central parts of the image. In the simulated projections, the modulation transfer function (MTF) and the finite size of the focal spot are neglected. If these factors were accounted for, the vertical structures would be less visible. Figures 3a-d show cutouts of section images central in the phantom (number 30 out of 64). The section image in 3a is reconstructed from simulated noise free primary projections and contains more visible structures, especially in lower mediastinum, compared to the two figures 3b (primary plus scatter) and 3c (primary plus scatter plus noise). In addition, the same structures are more visible in the noise free section image 3b compared to the noisy image in 3c. Figure 3c is similar to the section image reconstructed from the real phantom projections in 3d. Since the vertical artifacts, discussed above, appear at the same positions in all projection images, they are more visible in the reconstructed section images. The noise level is slightly overestimated in figure 3c. This could be due to inaccurate material composition data for the voxel phantom compared to materials in the real phantom, or other model uncertainties such as x-ray spectra. Nevertheless, this will be investigated in the future.

Future development

The model presented here for simulation of digital chest tomosynthesis is presently in an early stage of development. Since the current paper describes a work in progress, a future study is planned that solves some of the issues in this paper. A trial with human and model observers could be set up to evaluate the realism of the simulated section images compared to the real section images. Yet, due to the relatively coarse size of the voxels, the images cannot at the current state be

made indistinguishable. The problem with vertical artifacts in the images will partly be resolved if the blurring from the finite focal spot and the detector are taken into account in the model. In the analytical calculations of the primary projections, the spatial distribution of the focal spot and the resulting blurring can be accounted for by calculating several sub-projections with a slightly moved focus. Also, the detector blurring can be taken into account by filtering the images according to the detector MTF. A method for inclusion of simulated nodules will be implemented. The simulated images can then be used for trials with human and model observers to find dose efficient configurations.

The model presented here is intended for optimization of chest tomosynthesis. However, the simulations are currently rather time consuming which puts a limit on the number of systems that can be studied simultaneously with the present computer speed.

CONCLUSIONS

The current paper describes a work in progress that aims to developing a model for optimization of digital chest tomosynthesis. The simulated images are intended for trials with human and model observers to find dose efficient configurations. The method aims to allow for flexible variation, simulation and optimization of both existing and future chest tomosynthesis systems.

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Figure 1. Schematic view of the geometry used in the simulation model.

Figure 2. Tomosynthesis projections at 15 degrees: (a) real phantom image and (b) simulated image including contributions from scattered photons and added correlated noise.

Figure 3. Cutouts of tomosynthesis section images central in the phantom (section number 30 out of 64). The section images are reconstructed from (a) noise free primary projections, (b) primary projections including contributions from scattered photons and (c) projections as in (b) with added correlated noise. Figure 3d is the corresponding section image reconstructed from the real phantom projections.