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Feasibility of CT-based 3D anatomic mapping with a scanning-beam digital x-ray (SBDX) system

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Abstract

This study investigates the feasibility of obtaining CT-derived 3D surfaces from data provided by the scanning-beam digital x-ray (SBDX) system. Simulated SBDX short-scan acquisitions of a Shepp-Logan and a thorax phantom containing a high contrast spherical volume were generated. 3D reconstructions were performed using a penalized weighted least squares method with total variation regularization (PWLS-TV), as well as a more efficient variant employing gridding of projection data to parallel rays (gPWLS-TV). Voxel noise, edge blurring, and surface accuracy were compared to gridded filtered back projection (gFBP). PWLS reconstruction of a noise-free reduced-size Shepp-Logan phantom had 1.4% rRMSE. In noisy gPWLS-TV reconstructions of a reduced-size thorax phantom, 99% of points on the segmented sphere perimeter were within 0.33, 0.47, and 0.70 mm of the ground truth, respectively, for fluences comparable to imaging through 18.0, 27.2, and 34.6 cm acrylic. Surface accuracies of gFBP and gPWLS-TV were similar at high fluences, while gPWLS-TV offered improvement at the lowest fluence. The gPWLS-TV voxel noise was reduced by 60% relative to gFBP, on average. High-contrast linespread functions measured 1.25 mm and 0.96 mm (FWHM) for gPWLS-TV and gFBP. In a simulation of gated and truncated projection data from a full-sized thorax, gPWLS-TV reconstruction yielded segmented surface points which were within 1.41 mm of ground truth. Results support the feasibility of 3D surface segmentation with SBDX. Further investigation of artifacts caused by data truncation and patient motion is warranted.

Keywords

inverse geometry CT; scanning beam digital x-ray; iterative reconstruction; anatomic mapping

1. INTRODUCTION

Minimally invasive percutaneous transcatheter procedures have become common for a wide range of structural heart interventions including radiofrequency catheter ablation (RFCA) for atrial fibrillation and transcatheter aortic valve replacement. These procedures are performed using fluoroscopic imaging to facilitate device navigation to anatomic landmarks such as the pulmonary veins. X-ray fluoroscopy suffers from poor soft tissue contrast and as a result is often ill-suited to provide the necessary visualization of detailed anatomic features needed to perform certain transcatheter interventions. The integration of CT-derived 3D

anatomic maps can provide essential information for device placement. For example, during RFCA of cardiac arrhythmias the use of CT cardiac chamber models has been associated with superior patient outcomes including reduced arrhythmia recurrence and improved procedure efficacy.¹

Scanning-beam digital x-ray (SBDX) is a low-dose inverse geometry fluoroscopic technology capable of 3D localization of catheter devices.²⁻⁴ The purpose of this study was to investigate the feasibility of using SBDX rotational acquisition and CT reconstruction to generate 3D anatomic maps which could be paired with SBDX 3D device tracking. Simulated SBDX projection data is reconstructed using a statistical model-based iterative reconstruction method, a gridded filtered back projection method, and a gridded statistical iterative reconstruction method. High contrast objects representative of contrast-enhanced chambers are segmented from the CT images and the reconstruction methods are compared versus one another in terms of surface accuracy. The effects of noise level, incomplete angular sampling, and data truncation are examined in numerical simulations.

2. METHODS

2.1 Scanning-beam digital x-ray

The Scanning Beam Digital X-ray (Triple Ring Technologies, Inc; NovaRay Medical, Inc., Newark, CA) system is an inverse geometry fluoroscopy system designed for cardiac imaging (Figure 1).² SBDX utilizes an electromagnetically-scanned electron beam with a large-area transmission-style tungsten target. The electron beam is raster scanned over up to 100×100 focal spot positions on the target surface. The target is followed by a multi-hole collimator that defines a series of narrow x-ray beams convergent upon a CdTe photon-counting detector array. During fluoroscopic imaging the array of source positions is continuously scanned at 15 scan frame/s. The detector data captured during scanning is reconstructed into full field-of-view images in real time.

CT data acquisition can be achieved through simultaneous source scanning and C-arm rotation. In this paper we consider short-scan CT acquisitions using a 200 degree arc. For CT imaging, SBDX has a 14.6 cm diameter in-plane field-of-view (FOV) with 13.0 cm axial coverage when operating with the 71×71 source scan mode typical of cardiac imaging. If necessary, a maximum FOV of 19.3 cm by 17.7 cm can be obtained by operating with 100×100 source focal spots. Details of the SBDX geometry simulated in this study are summarized in Table 1

2.2 Reconstruction algorithms

Recently, novel inverse geometry CT (IGCT) systems similar to SBDX have been proposed for low-scatter CT imaging of large volumetric regions without cone-beam artifacts.⁵ Analytical reconstruction methods for IGCT include a gridded filtered backprojection approach,⁶ a Fourier rebinning algorithm,⁷ and a direct fan beam method with a non-uniform sampling correction.⁸ Iterative methods have been reported using a weighted least squares algorithm⁹ and an ordered subset expectation maximization algorithm.¹⁰

2.2.1 Statistical iterative reconstruction—Challenges with SBDX-based CT reconstruction include irregularly-organized and potentially high noise sinogram data resulting from a cardiac-gated short-scan acquisition. To accommodate these potential issues, a statistical model based iterative reconstruction method was investigated, in which reconstruction is cast as the optimization of a penalized weighted least-squares objective function.¹¹

$$\bar{x} = \arg \min_x \left[\frac{1}{2} (\mathbf{y} - \mathbf{Ax})^T \mathbf{D} (\mathbf{y} - \mathbf{Ax}) + \beta R(\mathbf{x}) \right] \quad (1)$$

Here $\mathbf{x}$ is the voxelized distribution of attenuation coefficient, $\mathbf{y}$ is measured projection data, $\mathbf{A}$ is the system matrix describing the intersection pathlength of each ray with each voxel, $R(\mathbf{x})$ is a regularization function, and $\mathbf{D}$ is a diagonal matrix used to assign relative weighting to data based on measurement statistics. The elements of $\mathbf{D}$ were set equal to the photon counts for a source-detector pair, and total variation (TV) was investigated as an example regularizer.

A nonlinear conjugate gradient method was used to minimize the objective function. The system matrix $\mathbf{A}$ was approximated using a ray-driven forward projector and the transpose operation was modeled using pixel-driven backprojection with linear interpolation at the detector. In this paper, reconstructions performed with $\beta > 0$ are termed PWLS-TV, and reconstructions with $\beta = 0$ are termed WLS.

2.2.2 Gridded filtered back projection—For comparison, the gridded filtered back projection (gFBP) method of Schmidt et al.⁶ was also implemented. The gFBP algorithm re-bins the IGCT projection data to a set of parallel ray projections that are regularly spaced in sinogram space. The gridded data are then reconstructed using filtered backprojection. The gridding parameters used are summarized in Table 2. The kernel widths were selected to be large enough that each of the SBDX projection rays was used and the output grid spacing was chosen to be as large as possible to minimize computation times while yielding acceptable reconstructions. A Hanning-windowed ramp filter was applied during filtered back projection.

2.2.3 Statistical iterative reconstruction with gridding—IGCT projection data sets can be large due to the multiple source elements. Consequently, statistical iterative reconstruction times may be limiting for intraprocedural CT imaging with SBDX. For example, the total number of projection rays within the SBDX-CT dataset is 1.16×10^{10} if operated in the 71×71 source scan mode with a 160×80 element detector array and 180 acquired view angles. The increased number of projection rays compared to conventional cone beam or parallel ray geometries results in increased computations for both forward and back projection. A voxel driven backprojector requires an iteration over the full image grid for each individual source point, which further increases back projection times by a factor of 71×71 for SBDX. However, gridding the projection data to a complete set of parallel ray projections (Table 2) reduced the size of the datasets used for this work by a factor of 524. Therefore, to investigate a more computationally efficient approach, we also applied the

statistical iterative reconstruction method of Eq. (1) to the gridded parallel ray projection data. In this case the system matrix $\mathbf{A}$ was adapted to describe the parallel ray geometry. The gridded iterative reconstructions are termed gPWLS-TV ($\beta > 0$) and gWLS ($\beta = 0$).

2.3 Simulations

An initial evaluation of reconstruction accuracy was obtained from a noise-free simulation of a high-contrast 3D Shepp-Logan phantom positioned at gantry isocenter. The phantom size was reduced to $80 \times 60 \times 60 \text{ mm}^3$ to avoid data truncation as a confounding factor. Line integral data was generated by analytical ray-tracing through the components of the phantom. The relative root mean square error (rRMSE) between the image and ground truth was calculated following reconstruction.

A thorax based on the FORBILD¹² phantom was simulated to evaluate the 3D surface accuracy of a segmented contrast-enhanced chamber in the presence of noise. Two phantom sizes were simulated in order to investigate performance without and with data truncation. The full sized thorax phantom contained a 3.6 cm diameter sphere positioned 2.4 cm anterior to isocenter. Sphere enhancement was set to +727 HU relative to background to simulate an arterial injection of iodinated contrast agent. A lower contrast 1.2 cm diameter cylinder (+200 HU) and resolution patterns (0.4, 0.5, 0.6, 0.7 mm) were also included. A reduced-size version of the thorax phantom ($100 \times 50 \times 60 \text{ mm}^3$) was generated by scaling all structure dimensions down by a factor of 2.4.

To simulate noisy projections, detector data was drawn from a Poisson distribution. The simulated source intensity (number of photons per source-detector element ray) was adjusted so that the mean number of counts exiting through the short axis of the phantom matched the mean photon counts measured on the SBDX system with a stack of acrylic in the beam. Three fluence levels were simulated, corresponding to 18.0, 27.2, and 34.6 cm thick acrylic phantoms imaged at 120 kV and full power (120 kVp, 200 mAp).

To simulate imaging of a more complicated 3D object, a 100 mm wide contrast-enhanced (+727 HU) phantom consisting of cuboids was constructed. The sharp edges of the cuboid phantom present a more challenging reconstruction and surface segmentation problem. The surface accuracy was evaluated as previously described.

Reconstructions were performed using a $512 \times 512 \times 385$ image grid with 0.234 mm^3 isotropic pixel resolution resulting in a 120 mm in-plane FOV and 90.2 mm axial coverage. All reconstruction algorithms were coded using the Common Unified Device Architecture (CUDA) programming model and C. Reconstruction was performed on a Windows PC with an Intel Xeon 3.60 GHz processor and an NVIDIA GeForce GTX 670 graphics processing unit.

2.4 Performance metrics

For each reconstruction method, background noise level, edge blurring, and chamber surface accuracy were evaluated. The noise level was quantified by measuring the standard deviation of image values (HU) in three regions-of-interest defined in the thorax phantom's heart, tissue and lungs. The edge blurring of the high contrast chamber was quantified by the

full width half maximum (FWHM) of a line spread function (LSF) derived from an angularly-averaged edge profile. Surface accuracy was calculated by performing intensity-based segmentation of the high contrast sphere, followed by a calculation of the distance of each segmented surface point to the closest point on the ground truth surface. The chamber surface was defined as the boundary of the voxels obtained after thresholding to half the intensity difference between the chamber and background regions. After calculating the distances to the ground truth surface, a histogram of distances was calculated and the 99th percentile distance was determined. The Sørensen-Dice coefficient was also calculated, in order to compare the segmented chamber volume with the ground truth volume.

2.5 Detector binning, scan range and gating

The SBDX system has a 320 column by 160 row detector that is normally operated in a 2×2 binning mode to produce 0.66 mm wide elements at the detector. To reduce noise in projection data and decrease computation time, the detector elements were further binned ($2 \text{ columns} \times 4 \text{ rows}$) prior to reconstruction for each of the methods examined. This resulted in a 1.32 mm in-plane pitch and 2.64 mm axial pitch (0.79 mm at isocenter). Following detector binning, the mean photon counts transmitted through the phantom were 52.2, 7.6 and 1.6 photons per ray for the three fluence levels simulated.

All CT simulations employed a short-scan consisting of 180 superviews uniformly distributed over 200 degrees. A superview refers to the detector data collected for one image frame period in fluoroscopic mode. When SBDX is operated in the 15 frame/s fluoroscopic mode with a 71×71 scan, each of the focal spot positions is visited by the electron beam 8 times in $1/15 \text{ s}$.² The 8 visits per focal spot are summed in the front-end of the image reconstructor.¹³

Although object motion was not simulated in this investigation, cardiac gating was simulated to explore the impact of incomplete sinogram data on image quality. To simulate gated acquisition, a subset of the superviews was extracted under the assumptions of a 33.3 degree/s gantry rotation speed, 60 bpm heart rate, and a 55% to 95% R-R gating window. This resulted in a total of 71 superview angles for a gated short-scan.

3. RESULTS

3.1 Noise free Shepp-Logan phantom

Figure 2 shows the noise-free miniature 3D Shepp-Logan phantom reconstructed using Eq. (1) with $\beta=0$ (no regularizer). Images are shown for transverse (axial) slices passing through isocenter and for planes superior and inferior to isocenter. The rRMSE for the full 3D reconstruction was 1.4%, demonstrating good agreement between the iterative reconstruction algorithm and ground truth. Figure 2D shows an intensity profile drawn vertically through the center of Figure 2A. Each of the small low contrast objects were fully resolved, as shown in Figure 2E.

3.2 Noise, resolution, and surface accuracy versus β

The β parameter in Eq. (1) controls the relative weighting between the data fidelity term and the total variation regularization function, with larger β values leading to increased noise suppression and smoothing. Figure 3 shows an example tradeoff between voxel noise standard deviation and spatial resolution, expressed as the full-width half-max (FWHM) of the linespread function. Results were obtained with the miniature thorax phantom and the lowest simulated fluence level (equivalent to 34.6 cm acrylic). Surface accuracy (99% percentile method) and Sørensen-Dice coefficient are plotted versus β parameter in Figure 4A and 4B, respectively.

The curve in Fig. 3 corresponds to the gridded version of PWLS-TV (gPWLS-TV). Points for the gFBP, WLS (i.e. $\beta=0$), and PWLS-TV methods are also shown. As a practical matter, it was not feasible to generate full curves for non-gridded 3D PWLS-TV due to the long reconstruction time involved. Example reconstruction times for the different methods are shown in Table 3. In the interventional setting, reconstruction time is an important consideration. Timing results indicated that the most promising methods for further investigation would involve gridding. Consequently, the remainder of this paper focuses on gFBP, gWLS, and gPWLS-TV. For the gPWLS-TV method, a β value of 0.1 was found to maximize surface accuracy and the Sørensen-Dice coefficient (Fig 4). Since the focus of this work is the segmentation of large high contrast objects, this β value was assumed to be most suitable for tasks involving segmentation of large high contrast objects, and was used for the reconstructions reported below.

3.3 Miniature thorax phantom

Example transverse slices of the reduced size 3D thorax phantom are shown in Figure 5, for three fluence levels and three reconstruction methods: i) gridded FBP, ii) gridded WLS and iii) gridded PWLS with TV regularization ($\beta=0.1$). Figure 6 shows reconstructed sagittal slices under the same conditions. For the three fluence levels, the mean photon counts transmitted through the phantom were 52.2, 7.6 and 1.6 photons per ray (after the detector binning described in Sec. 2.5). Gating was not applied to the data used for these reconstructions.

Qualitatively, the gFBP and gWLS have similar appearance, while the gPWLS-TV has reduced noise and slightly less sharp edges. The reduction in sharpness for gPWLS-TV is a consequence of the choice in β parameter, which was optimized for surface accuracy (Sec 3.2). These observations are reflected in the quantitative measures of noise, spatial resolution, and surface accuracy, presented in Table 4. The gPWLS-TV method had the lowest voxel noise, measuring -66% , -61% , and -53% relative to gFBP for the three fluence levels. Averaging the full-width half-maximum of the LSF across fluence levels, the results for gFBP, gWLS, and gPWLS-TV were 0.96 mm, 1.03 mm, and 1.25 mm, respectively.

Under the highest noise condition (34.6 cm acrylic equivalent), the surface accuracy metrics show that the gPWLS-TV method yielded the most accurate segmentations. In this case, 99% of the segmented boundary points were within 0.70 mm of the ground truth surface,

versus 1.20 mm and 1.35 mm for the gFBP and gWLS methods. In the lower noise conditions (18.0 cm, 27.2 cm acrylic equivalent) there was a slight improvement in surface accuracy for gPWLS-TV and gWLS relative to gFBP, although results were generally similar (e.g. 0.33–0.41 mm in the lowest noise case). The trends in the Sørensen-Dice coefficient followed those observed in the 99th percentile deviation. Overall, results suggest that, for segmentation tasks in an interventional setting, the benefits of iterative reconstruction with regularization are mainly seen in low flux imaging scenarios. For high flux imaging, the gFBP method may be more expedient.

3.4 Cuboid phantom

The cuboid phantom was reconstructed for each of the three fluence levels using gFBP and gPWLS-TV in order to evaluate surface accuracy for a more complicated object (Figure 7). The cuboid images were segmented using intensity based thresholding. The 3D surface models were volume rendered using ImageVis3D (NIH/NIGMS).

Surface accuracy results for the cuboid phantom are summarized in Table 5. For the highest fluence case, 99% of perimeter points were within 0.47 mm of ground truth for both the gFBP and gPWLS-TV reconstructions. For the second highest fluence case, gPWLS-TV slightly outperformed gFBP with 99% of perimeter points being within 0.70 mm of ground truth versus 1.17 mm for gFBP. The Sørensen-Dice coefficient exceeded 0.95 for each of the four reconstructed surfaces at the two highest fluences. For the lowest fluence analyzed, 99% of perimeter points were within 1.88 mm of the ground truth for gPWLS-TV versus 2.82 mm for gFBP. The sharp edges of the cuboid phantom were challenging to reconstruct under high noise conditions and resulted in increased maximum surface errors of 4.54 mm (gPWLS-TV) and 7.05 mm (gFBP). Interestingly, the Sørensen-Dice coefficient for gPWLS-TV reconstruction measured 0.93, indicating agreement with the ground truth surface volume. The Sørensen-Dice coefficient for the gFBP case was 0.79.

3.5 Large truncated thorax phantom

The effects of data truncation were investigated by performing 3D reconstructions of the large thorax phantom with the highest fluence level (18.0 cm acrylic equivalent). Figure 8 demonstrates transverse slices of the truncated thorax phantom using gPWLS-TV and gFBP. As expected, the truncated reconstructions demonstrated bright shading artifacts toward the periphery and inaccurate CT numbers. Despite this, intensity-based segmentation of the high contrast chamber was successful. The gFBP method showed improved surface accuracy versus gPWLS-TV with 99% of the segmented surface points being within 0.58 mm of the ground truth versus 1.00 mm for gPWLS-TV. Note the β value (0.1) used for the gPWLS-TV reconstruction was optimized for the lowest fluence case, which resulted in increased edge blurring and decreased image noise versus gFBP. The LSF FWHM of gPWLS-TV measured 1.22 mm versus 0.99 mm for gFBP. The noise standard deviation within the high contrast chamber region measured 17.3 HU for gPWLS-TV versus 31.9 HU for gFBP. The Sørensen-Dice coefficient measured 0.98 for both reconstruction methods.

3.6 Gated and truncated projections

Figure 9 demonstrates 2D gPWLS-TV and gFBP reconstruction of the full sized thorax phantom using gated projection data (18.0 cm acrylic equivalent fluence). The view angle undersampling created by gating caused streak artifacts in both images. For the gPWLS-TV case, the high contrast chamber could still be recovered by applying a gradient-based segmentation scheme which is less sensitive to gradual brightness variations. The surface segmented from the gPWLS-TV reconstruction was within 1.41 mm of the ground truth 99% of the time. The gFBP surface was not successfully segmented by simple intensity-based or gradient-based methods due to the level of artifacts present.

4. DISCUSSION

This work investigates the feasibility of 3D anatomic surface mapping via CT reconstruction of projection data acquired with an inverse geometry C-arm fluoroscopic system. Iterative reconstruction methods and an existing FBP method were evaluated for three different noise levels, with and without data truncation and view angle undersampling. The different reconstruction methods were compared with a specific focus placed on the segmentation accuracy of 3D high contrast chambers.

Because of the multi-source array, SBDX can generate projection datasets that are an order of magnitude or larger in size than third generation CT systems. This is an important consideration for intraprocedural CT image reconstruction since reconstruction time may limit workflow. The 3D WLS and PWLS-TV reconstruction methods presented in this work required long reconstruction times despite a parallel implementation using the CUDA programming model. Further optimization of reconstruction time may be possible. However in this work, gridded iterative reconstruction was investigated as a practical means to reduce computation times while retaining the features of statistical iterative reconstruction. Specifically, we are interested in methods to deal with potentially high noise projection data and incomplete sinogram data. Gridded PWLS-TV reduced the reconstruction time by a factor of 293 compared to the nongridded PWLS method.

The gFBP and gPWLS-TV reconstruction methods provided very comparable surface accuracy for the two highest fluences analyzed in this work. For lower fluence imaging scenarios, there was a benefit to using gPWLS-TV. For example, reconstruction of the cuboid phantom at the lowest fluence yielded 1.88 mm surface accuracy for gPWLS-TV versus 2.82 mm for gFBP. The improvement may be attributed to the noise suppression obtained through regularization.

Results support the feasibility of accurate anatomic surface mapping with the SBDX system, however several additional challenges must be addressed. The small field-of-view results in projection data truncation that leads to shading artifacts, which, if unaccounted for during image reconstruction, increase the difficulty of segmentation. Angular undersampling in a gated acquisition increases image noise and causes reconstruction artifacts that reduce surface accuracy. Gated and truncated datasets were briefly investigated here for a stationary object. Future work should also examine the effects of cardiac motion on surface accuracy.

Reconstruction techniques which address data truncation and gating-induced artifacts would help improve the accuracy and stability of 3D surface mapping results. Prior image constrained compressed sensing (PICCS)¹⁴ is a reconstruction method that is capable of generating time resolved cardiac images for a slowly rotating C-arm system¹⁵ in the presence of data truncation.¹⁶ Future work will include implementing and evaluating a PICCS reconstruction algorithm for SBDXCT, and optimization of gridding parameters for surface accuracy. Finally, the findings in simulations should be confirmed experimentally with the SBDX prototype system.

5. CONCLUSIONS

Segmentation of high contrast objects from an SBDX rotational acquisition is feasible using inverse geometry CT reconstruction. Iterative reconstruction of gridded projection data reduces reconstruction times relative to conventional iterative reconstruction. Surface accuracy of gridded FBP, and gridded PWLS-TV methods are similar in nongated acquisitions at high fluence levels, however gridded PWLS-TV offers improved accuracy at low fluence. Iterative techniques can be adapted to irregular sinogram data in gated acquisitions. Future work will examine more complex patient geometries, effects of object motion, and techniques to improve image quality in truncated and gated acquisitions. These initial simulations are an important first step towards developing a 3D anatomic mapping capability for the SBDX system.

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Figure 1.

SBDX uses a raster scanned focal spot. CT imaging is possible with simultaneous scanning and a C-arm rotation.

Figure 2.

The reduced-size 3D Shepp-Logan phantom reconstructed using the weighted least squares algorithm. Transverse slices through isocenter (A) and at 6.6 mm off axis (B). A sagittal slice is shown (C). A vertical profile through (A) is plotted versus the reference profile in (D). A magnified view of the low contrast objects of the phantom is shown in (E).

Figure 3.

A noise-resolution tradeoff curve is plotted for different β values for the miniature thorax phantom at the lowest simulated photon fluence.

Figure 4. The surface accuracy (A) and Sørensen-Dice coefficient (B) versus β value, for the miniature thorax phantom at the lowest simulated photon fluence. The dashed line represents the surface accuracy of a gridded FBP reconstruction.

Figure 5.

Axial slice reconstructions of the reduced-size thorax phantom using gFBP (top row), gWLS (middle row) and gPWLS-TV with $\beta=0.1$ (bottom row) reconstruction. The mean fluence transmitted through the phantoms are equivalent to imaging through 18.0, 27.2, and 34.6 cm acrylic (left to right). Display window is $[-800, 800]$ HU.

Figure 6. Sagittal slice reconstructions of the reduced-size thorax phantom using gFBP (top row), gWLS (middle row) and gPWLS-TV with $\beta=0.1$ (bottom row) reconstruction. The mean fluence transmitted through the phantoms are equivalent to imaging through 18.0, 27.2, and 34.6 cm acrylic (left to right). Display window is $[-800, 800]$ HU.

Figure 7. Reconstructions of the cuboid phantom using gFBP (top row) and gPWLS-TV (middle row). The gPWLS-TV reconstructions were segmented and the high contrast chamber was volume rendered (bottom row). The mean fluence transmitted through the phantoms are equivalent to imaging through 18.0, 27.2, and 34.6 cm acrylic (left to right). Display window is $[-800, 800]$ HU.

Figure 8. Reconstruction of the full size thorax phantom with 3D gridded PWLS-TV (A) and 3D gFBP (B) methods. The mean fluence transmitted through the phantom was equivalent to 18.0 cm of acrylic. Display window is $[-200, 1000]$ HU.

Figure 9.

Reconstruction of gated data from the full sized thorax phantom, for 2D gridded PWLS-TV (A) and gFBP (B). The display windows were set to $[-200, 1000]$ (A) and $[-600, 800]$ (B) in order to emphasize the reconstruction artifacts. The mean fluence through the phantom was equivalent to imaging through 18.0 cm of acrylic.

Table 1**SBDX system geometry simulated**

Source-detector-distance	1500 mm
Source-isocenter-distance	450 mm
Number of focal spot positions	71×71
Focal spot pitch	$2.3 \text{ mm} \times 2.3 \text{ m}$
Native detector array	320×160
Native detector element pitch	0.33 mm
Gantry angles per shortscan	180
Shortscan angular range	200 degrees

Table 2

Parameters used with gFBP method

Number of view angles over 180 degrees	240
Rays per 2D view [columns $\times$ rows]	240 $\times$ 385
2D view sampling pitch	0.5 mm $\times$ 0.23 mm
Radial kernel	1.4 mm $\times$ 2.8 mm
Angular kernel	1 degree $\times$ 1 degree

Table 3

Average 3D reconstruction times

	Minutes
WLS	21,600.0
PWLS-TV	29,170.5
gFBP	0.1
gWLS	88.1
gPWLS-TV	83.1
Gridding time	16.2

Reconstructed surface accuracy, LSF FWHM, and noise standard deviation for three fluence levels and three reconstruction methods.

Table 4

	Surface Accuracy 99th Percentile (mm)	Sørensen-Dice Coefficient	LSF FWHM (mm)	Noise Standard Deviation (HU)		
				Heart	Tissue	Lung
<i>18.0 cm acrylic fluence</i>						
gFBP	0.41	0.97	0.92	27.9	24.8	17.9
gWLS	0.33	0.96	1.06	19.7	19.1	13.5
gPWLS-TV	0.33	0.95	1.22	9.2	10.5	5.0
<i>27.2 cm acrylic fluence</i>						
gFBP	0.53	0.96	0.94	64.5	65.7	47.1
gWLS	0.47	0.96	1.05	45.4	47.6	36.7
gPWLS-TV	0.47	0.95	1.23	23.4	24.4	20.0
<i>34.6 cm acrylic fluence</i>						
gFBP	1.20	0.89	1.01	189.5	128.2	104.3
gWLS	1.35	0.86	0.98	217.0	150.3	123.3
gPWLS-TV	0.70	0.94	1.29	92.3	52.9	52.5

Table 5

Surface accuracy results of a cuboid phantom reconstructed using gFBP and gPWLS-TV for three fluence levels.

	Surface Accuracy 99th Percentile (mm)	Max Surface Deviation (mm)	Sørensen- Dice Coefficient
<i>18.0 cm acrylic fluence</i>			
gFBP	0.47	0.94	0.98
gPWLS-TV	0.47	0.70	0.98
<i>27.2 cm acrylic fluence</i>			
gFBP	1.17	2.58	0.96
gPWLS-TV	0.70	1.64	0.98
<i>34.6 cm acrylic fluence</i>			
gFBP	2.82	7.05	0.79
gPWLS-TV	1.88	4.54	0.93