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Localization of cardiac volume and patient features in inverse geometry x-ray fluoroscopy

Michael A. Speidel^{a,b}, Jordan M. Slagowski^a, David A.P. Dunkerley^a, Martin Wagner^a, Tobias Funk^c, and Amish N. Raval^b

^aDept. of Medical Physics, University of Wisconsin, Madison, WI, USA

^bDept. of Medicine, University of Wisconsin, Madison, WI, USA

^cTriple Ring Technologies, Inc., Newark, CA, USA

Abstract

The scanning-beam digital x-ray (SBDX) system is an inverse geometry x-ray fluoroscopy technology that performs real-time tomosynthesis at planes perpendicular to the source-detector axis. The live display is a composite image which portrays sharp features (e.g. coronary arteries) extracted from a 16 cm thick reconstruction volume. We present a method for automatically determining the position of the cardiac volume prior to acquisition of a coronary angiogram. In the algorithm, a single non-contrast frame is reconstructed over a 44 cm thickness using shift-and-add digital tomosynthesis. Gradient filtering is applied to each plane to emphasize features such as the cardiomeastinal contour, diaphragm, and lung texture, and then sharpness vs. plane position curves are generated. Three sharpness metrics were investigated: average gradient in the bright field, maximum gradient, and the number of normalized gradients exceeding 0.5. A model correlating the peak sharpness in a non-contrast frame and the midplane of the coronary arteries in a contrast-enhanced frame was established using 37 SBDX angiographic loops (64–136 kg human subjects, 0–30° cranial-caudal). The average gradient in the bright field (primarily lung) and the number of normalized gradients >0.5 each yielded peaks correlated to the coronary midplane. The rms deviation between the predicted and true midplane was 1.57 cm. For a 16 cm reconstruction volume and the 5.5–11.5 cm thick cardiac volumes in this study, midplane estimation errors of 2.25–5.25 cm were tolerable. Tomosynthesis-based localization of cardiac volume is feasible. This technique could be applied prior to coronary angiography, or to assist in isocentering the patient for rotational angiography.

Keywords

inverse geometry; x-ray fluoroscopy; scanning-beam digital x-ray; tomosynthesis; interventional cardiology

1. INTRODUCTION

The x-ray angiographic system is an integral part of diagnostic and therapeutic procedures performed in the cardiac catheterization laboratory, providing the high spatio-temporal resolution and wide anatomic coverage needed to visualize catheter-based devices and patient anatomy. Although the importance of x-ray imaging in the treatment of coronary

artery disease and structural heart disease is well-established, radiation dose to the patient and to the staff via x-ray scatter are ongoing sources of concern.^{1,2} Additionally, the lack of depth resolution in an x-ray projection presents a challenge in structural heart interventions involving device navigation within relatively large and complex 3D spaces such as the cardiac chambers.³ These limitations have become increasingly apparent as cardiac interventional procedures have grown in complexity.

Scanning-beam digital x-ray (SBDX) is an inverse geometry fluoroscopic/angiographic technology designed to both improve dose efficiency and provide depth resolution in cardiac interventional procedures.⁴⁻⁶ In contrast to a conventional system that images the patient volume with a wide cone beam, SBDX has a spatially-distributed x-ray source that performs a high speed narrow-beam scan of the patient volume for every frame period. As shown in Figure 1, an electron beam inside the x-ray tube performs a raster scan across an array of focal spot positions defined on a transmission-style tungsten target. A multihole collimator just beyond the target defines a series of x-ray beamlets, each of which is directed at a high speed, small area, detector array. The raw detector images captured from each scan frame are reconstructed into a full-field-of-view image for real-time fluoroscopic display. This is termed an inverse geometry fluoroscopy because the collection of x-ray beamlets form an overall x-ray field that is largest on the source side and smallest on the detector side.

Inverse geometry fluoroscopy is designed to achieve a several-fold dose reduction by i) reducing image-degrading scatter through the use of a narrow x-ray beam and large air gap, ii) using a geometry that distributes primary x-rays over a larger area at the skin entrance, and iii) using a 2-mm thick CdTe photon counting detector that maintains high detective quantum efficiency at high tube voltage.⁶ Since inverse geometry fluoroscopy provides a range of view angles for each point in the patient volume in each frame period (see Fig. 1), it also represents a form of real-time tomosynthesis. Real-time tomosynthesis offers novel capabilities that are unavailable with conventional fluoroscopy. For example, it has been shown that the depth resolution inherent to SBDX tomosynthesis can be exploited to perform real-time 3D catheter tracking,^{7,8} 3D vessel analysis,⁹ stereoscopic fluoroscopy,¹⁰ and lung tomosynthesis.¹¹ However, inverse geometry fluoroscopy also requires a different approach to forming a 2D display analogous to conventional fluoroscopy.

The SBDX system forms a live 15 frame/s fluoroscopic image display using a real-time reconstructor.^{4,5} For each scan frame, a stack of tomosynthesis planes spanning the target anatomy is reconstructed (Figure 2(a)). A pixel-by-pixel plane selection algorithm is then applied to extract pixels corresponding the high contrast and sharp features located in the plane stack. The resulting image, termed a multiplane composite, is analogous to a conventional display (Fig 2(b)). The current SBDX prototype uses a reconstruction volume that is 16 cm thick along the source-detector direction, centered on gantry isocenter by default. While this provides adequate coverage for the heart, currently the heart must be manually isocentered. Isocentering can be accomplished by sequentially imaging from two gantry angles (e.g. posterior-anterior and lateral) and adjusting the table height to ensure that the cardiac silhouette or a cardiac catheter is centered in both views. Clinically, it would be preferable to have an automatic method for predicting the position of the cardiac volume from a single frame of imaging, prior to contrast-enhanced angiography. This information

could be used to adjust the position of the reconstruction volume, or alternatively, set the table height.

The purpose of this work was to develop a localization algorithm that estimates of the position of the cardiac volume along the source-detector axis so that the reconstruction volume may be properly centered. The algorithm uses SBDX tomosynthesis to automatically localize natural anatomic features of the lung, heart, and diaphragm from an image frame acquired without iodine contrast. A model is used to map these feature positions to the location of the cardiac volume. Algorithm performance was evaluated with SBDX data acquired in human subjects.

2. METHODS

2.1 SBDX system

The SBDX system consists of an x-ray tube with an electronically scanned focal spot, a multi-hole collimator, a high speed photon counting detector array, and a GPU-based real-time image reconstructor. The current system has 100 x 100 focal spot positions on a 2.3 mm pitch and a 5.3 cm x 10.6 cm photon-counting detector array (160 x 320 elements) positioned 150 cm from the target plane in the x-ray tube.⁵ The detector converter material is 2 mm thick CdTe. The tomographic angle, defined as the angular span of x-rays passing through a point at isocenter, is 6 degrees in the axial plane and 3 degrees in the sagittal plane. For cardiac imaging, 71 x 71 focal spot positions are scanned in each 1/15 s frame. In this scan mode, the electron beam visits each focal spot position 8 times per frame, resulting in $71 \times 71 \times 8 = 40,328$ detector images per frame.

The real-time reconstructor has 8 GPUs which perform shift-and-add tomosynthesis reconstruction (i.e. unfiltered back projection) at 32 programmable plane positions spaced by 5 mm (16 cm thick volume). In each image, in-plane objects appear sharp and out-of-plane objects are progressively blurred according to their distance from the reconstruction plane. The 5 mm plane spacing is set to ensure that features residing between planes experience minimal out-of-plane blurring, relative to other blurring mechanisms in the system (e.g. focal spot, detector element).⁴ The live 15 frame/s composite display is formed by selecting the plane of highest contrast and sharpness for each display pixel position, using an algorithm tuned for coronary angiography. Details can be found in Refs. 4 and 5.

Example tomosynthesis planes reconstructed from a single 1/15 s scan frame of a coronary angiographic sequence and the corresponding composite image are shown in Figures 2(a) and 2(b). The stack of 32 tomosynthesis images shows different planes of focus within the coronary arterial tree. Since the plane stack encompasses the cardiac volume, the plane selection algorithm is able to select in-focus planes for all points along the coronary arteries (Fig 2(b)). Figure 2(c) shows the possible effect of an improperly centered reconstruction volume. In this case, the reconstruction volume is shifted towards the x-ray source by 5 cm, such that an anterior coronary artery falling outside the reconstruction volume is portrayed with reduced contrast and sharpness in the composite image.

2.2. Cardiac volume localization algorithm

The problem of localizing the cardiac volume arises from the fact that it is computationally impractical to continuously reconstruct a large volume that is guaranteed to encompass the patient, in real-time, while maintaining the necessary spatial resolution. With a restricted volume, it is necessary to judge where coronary arteries will be located *prior* to the injection of contrast agent. To solve this problem, we propose to measure image sharpness versus tomosynthesis plane position from a single non-contrast scan frame using a process that is similar to autofocus in digital photography.¹² It is expected that a curve describing tomosynthesis image sharpness vs. plane position (z) will exhibit maxima at z -planes corresponding to the edges of natural anatomical features, such as the vertebrae, ribs, diaphragm boundary, lung vasculature, and cardiomedastinal contour. Furthermore, the positions of these anatomic features are expected to be correlated to the position of the cardiac volume along the source-detector axis. Therefore, the localization algorithm consists of four steps: 1) tomosynthesis reconstruction over a wide range of plane positions for a single frame, 2) filtering of tomosynthesis images to emphasize natural anatomic features, 3) depth localization of features along the source-detector axis, and 4) mapping of the localized feature position to the midplane of the cardiac volume.

2.2.1 Tomosynthesis reconstruction—In the first step, a single frame of fluoroscopy is reconstructed at a wide range of planes encompassing the full thickness of the patient chest. Standard SBDX shift-and-add tomosynthesis is used, as described in Sec 2.1. The resulting set of images is analogous to those that would be obtained in conventional photography by varying the focus distance. The tomosynthesis image planes $T(i,j,k)$ are reconstructed at predefined distances z_k (mm) from the target plane of the x-ray tube, where indices i,j refer to pixel row and pixel column, and k is the plane index. Since patient size and location are not known *a priori*, the thickness of the reconstruction volume along the source-detector axis must be set conservatively. In this study, a 44 cm thick volume (88 planes x 5 mm) centered on gantry isocenter was sufficient for 0 to 30° cranial-caudal imaging of 64–136 kg subjects.

2.2.2 Feature emphasis in images—In the second step, each plane is filtered to emphasize contrast and natural edges in the patient anatomy. A number of filtering methods are possible. For this work, we chose a framework similar to one that was recently implemented in real-time for SBDX 3D catheter tracking.⁸ First, each image plane is low-pass filtered by a small kernel h_1 to reduce image noise:

$$F_1(i, j, k) = T(i, j, k) * h_1(i, j) \quad (1)$$

where $T(i,j,k)$ is the stack of tomosynthesis planes from step 1 (Sec 2.2.1). Next, to facilitate the calculation of image gradients which are normalized to background intensity, each image is divided by a highly smoothed version of itself. Using subscript “1n” to represent the normalized version of F_1 ,

$$F_{1n}(i, j, k) = F_1(i, j, k) / [F_1(i, j, k) * h_n(i, j)] \quad (2)$$

where h_n is the smoothing kernel and the division is performed elementwise. To estimate edge sharpness in each plane, the mean of the absolute image gradients is computed from the background normalized images,

$$G_1(i, j, k) = \frac{1}{2} (|F_{1n} * g_x(i, j)| + |F_{1n} * g_y(i, j)|) \quad (3)$$

where g_x and g_y are extended Prewitt kernels that estimate the gradient along the image columns and rows, respectively.¹³ Finally, additional low pass filtering with a kernel h_2 is applied to the background-intensity-normalized gradient values in each plane:

$$G_2(i, j, k) = G_1(i, j, k) * h_2(i, j) \quad (4)$$

The kernel h_2 reduces gradient noise and also serves to combine edge values on the perimeter of discrete features. In this study the following convolution kernel sizes were used: $h_1 = 3 \times 3$ pixel boxcar, $h_n = 100 \times 100$ pixel boxcar, $h_2 = 3 \times 3$ pixel boxcar. All kernels were normalized to unit area.

2.2.3 Depth localization of features—The third step, depth localization of features, is based on a metric of sharpness versus plane position z calculated from the filtered tomosynthesis images $G_2(i, j, k)$. Generally, this may be represented by an operator $f(\cdot)$ that reduces the gradient information in the k -th plane to a scalar sharpness metric $s(k)$

$$s(k) = f(G_2(i, j, k)) \quad (5)$$

Empirically, it is observed that lung fields are associated with a large number of modest-sized gradients (after normalization by background intensity), whereas the diaphragm and cardiac contours produce high gradients at relatively few pixels in the image scene (i.e. there is high edge contrast along the contours only). Thus, for the operator $f(\cdot)$, we considered i) the average of the image gradient values, ii) the number of image gradient values exceeding a threshold, and iii) the maximum image gradient value.

A regional sharpness metric which only analyzed the gradient values located within the bright areas of the input images T was also considered in order to improve the sensitivity of the sharpness metric to lung features. The threshold intensity t_{bright} corresponding to the bright field was defined as the 2nd threshold obtained from a 3-threshold version of Otsu's multithreshold algorithm,¹⁴ when applied to a maximum intensity projection of $T(i, j, k)$ along the k dimension (source-detector axis). Given this threshold value, a binary mask $B(i, j, k)$ is constructed with 1's corresponding to image positions of T with intensity

exceeding t_{bright} and then Eq. (5) is calculated considering only the pixel positions of $G_X(i,j,k)$ where $B(i,j,k) = 1$.

Figure 3 shows example sharpness vs. plane position curves from the same image frame using three different sharpness metrics: i) average gradient value within the bright field of the image, ii) the number of pixels in the entire gradient image which exceed 0.5 (i.e. 50% threshold), and iii) the maximum gradient value in the entire gradient image. Also shown are selected tomosynthesis images T and gradient images G_2 from the plane stack. The boundary of the bright field in the upper right quadrant is outlined in green. In this example, the average gradient in the bright field produced a broad sharpness curve with a single peak, corresponding mainly to where lung vasculature/airway was in best focus. In contrast, using the global number of gradient values exceeding a threshold or the global maximum gradient resulted in sharpness curves with multiple tighter peaks. The maximum gradient approach was most sensitive to high contrast features. As shown in Fig. 3, the maximum gradient approach produced peaks corresponding to sternal wires ($z=51$ cm), cardiomedistinal contour and the diaphragm border ($z=44.5$ cm), lung texture and guide catheter ($z=38$ cm), and posterior ribs ($z=33$ cm).

The localized position of a feature can be the position of the sharpness peak, or alternatively, the centroid of the peak above some threshold value, as in Ref. 7. The centroid approach has the advantage that localization resolution is not limited to the plane-to-plane spacing. However in this study, for simplicity, the peak position was used. Noting that the tomosynthesis images $T(i,j,k)$ are reconstructed at predefined distances z_k (mm) from the x-ray tube, the sharpness metric $s(k)$ can be easily re-expressed as a function of distance z , i.e. $s(k) \rightarrow s(z_k)$, and the localized position along the z -axis is given by:

$$z_{feature} = \arg\max_{z_k} s(z_k) \quad (6)$$

2.2.4 Mapping of localized features to cardiac volume position—A sharpness metric that reflects the positions of anatomic features of the heart, lungs, and bones along the source-detector axis is expected to be correlated to the midplane of the cardiac volume. A statistical model mapping the localized features to the cardiac volume was established using 37 fluoroscopic/angiographic loops from 19 consecutively screened patients who required clinically indicated diagnostic coronary angiography. Human subjects institutional review board approval was obtained for this prospective study. Each subject had their standard of care angiogram performed on a conventional system. The patient table was then rotated away from the conventional system to the SBDX system in the same clinical catheterization room with the patient remaining supine. Two SBDX angiograms were acquired at the same projection angles, catheter position, and contrast rate and volume as conventional imaging.

The ground truth position of the cardiac volume midplane was established for each loop by reconstructing and visually inspecting a wide (88-plane) stack of an iodine-enhanced frame of the coronary angiogram. An offline GPU-accelerated version of the SBDX real-time reconstruction algorithm was used to facilitate reconstruction of the wide plane stacks. The

true coronary mid-plane was defined as the average of the highest (z_{max}) and lowest (z_{min}) planes containing a visually in-focus coronary artery:

$$z_{heart,true} = (z_{max} + z_{min}) / 2 \quad (7)$$

To establish the mapping model, wide plane stacks, sharpness curves, and feature positions were generated from an early non-contrast frame of each angiographic loop, as described in Secs. 2.2.1 – 2.2.3. An l_1 -penalized linear regression (LASSO) analysis was performed to establish a statistical model relating $z_{feature}$ measurements and $z_{heart,true}$ positions and also to identify the sharpness metrics which have the greatest predictive value. Specifically, the standard LASSO problem¹⁵

$$(\hat{\beta}_0, \hat{\beta}) = \underset{\beta_0, \beta}{\operatorname{argmin}} \frac{1}{2} \|y - \beta_0 - X\beta\|_2^2 + \lambda \|\beta\|_1 \quad (8)$$

was formulated, with the $z_{heart,true}$ values from all loops arranged in the $n \times 1$ column vector y , and column vectors of $z_{feature}$ values, each generated by a different sharpness metric, arranged in the $n \times p$ matrix $X = [x_1 \ x_2 \ \dots \ x_p]$. The $p \times 1$ column vector β contains the weighting coefficients for each sharpness metric, and the scalar β_0 is an intercept value. The regularizer term shrinks or sets to zero some of the weighting coefficients, therefore encouraging a solution with only meaningful sharpness metrics. LASSO was implemented with MATLAB R2013a (The MathWorks Inc., Natick, MA). The regularization strength λ was set to the largest value that gave a non-null model.

The solution to Eq. (8) yields weighting coefficients for the model that predicts the position of the coronary midplane from localized features. In practice, for some new image frame, the predicted coronary midplane is

$$z_{heart} = x\hat{\beta} + \hat{\beta}_0 \quad (9)$$

where the row vector $x = [z_{feature,1}, z_{feature,2}, \dots, z_{feature,p}]$ contains the set of $z_{feature}$ values generated by different sharpness metrics for the image frame.

2.3 Cross validation

The performance of the cardiac volume localization algorithm was evaluated by leave-one-out cross validation with the 37 angiographic loops described in Sec 2.2.4. That is, for each loop, i) a model was generated according to Eq. (8) leaving the loop of interest out of the model generation process, ii) the feature positions $x = [z_{feature,1}, z_{feature,2}, \dots, z_{feature,p}]$ were calculated for the one left-out loop, iii) the predicted z_{heart} value was calculated by Eq. (9), and iv) the result was compared to the ground truth $z_{heart,true}$ value for that same loop. The deviations from ground truth $z = (z_{heart} - z_{heart,true})$ obtained by leave-one-out cross

validation were summarized using the root mean squared deviation (RMSD). Given n (37) deviations,

$$RMSD = \sqrt{\frac{1}{n} \sum_{i=1}^n \Delta z_i^2} \quad (10)$$

3. RESULTS

Sharpness versus plane position curves were based on i) the average gradient in the bright lung field, ii) the number of gradients exceeding 0.5 in the full image, and iii) the maximum gradient in the image. Figure 4 compares the peak positions of these three metrics measured from an early non-contrast enhanced fluoroscopic frame to the ground truth coronary midplane ($z_{heart,true}$) determined from a later contrast-enhanced angiographic frame, for each of the 37 recorded loops. Subject AP chest thickness was 22 to 32 cm, weight was 64–136 kg, and imaging was performed at 0 to 30° cranial-caudal angulation. Both average gradient in the bright field and the number of gradients > 0.5 yielded sharpness curves with stable peaks correlated to the coronary midplane.

Sharpness measured in terms of the maximum gradient in the image was sensitive to a number of anatomical structures, as demonstrated in Fig. 3, however the position of the strongest peak was not always consistent from one subject to the next (see Fig. 4). Specifically, when vascular access was obtained via the femoral artery there was a high contrast guide catheter in the thoracic aorta running vertically across the image, but when radial artery access was used, this catheter was not as visible in the x-ray image. Furthermore, some but not all patients in this study had sternal wires from prior cardiac surgery which produced very strong sharpness peaks. Due to inter-subject variability in these types of features, the plane of the maximum gradient was not as useful for predicting cardiac volume position.

The model for predicting the coronary midplane was:

$$z_{heart} = 0.473z_1 + 0.217z_2 - 0.019z_3 + 20.14 \text{ [cm]}$$

where z_1 = the peak location of the average gradient in the bright field, z_2 = the peak location of the number of gradients exceeding 0.5, z_3 = the peak location of the maximum gradient, and all distances are measured relative to the target plane of the x-ray source in units of cm. In leave-one-out cross validation, the root mean square error between the predicted and true coronary midplane position was 1.60 cm. The very small weighting coefficient assigned to the maximum gradient indicated that this sharpness metric could be excluded. In this case the model was:

$$z_{heart} = 0.484z_1 + 0.175z_2 + 20.45 \text{ [cm]}$$

and the cross-validation root mean squared deviation was 1.57 cm.

The reconstruction volume of the current SBDX prototype is 16 cm thick, and in this study the coronary arteries occupied volumes ranging from 5.5 cm to 11.5 cm thick as determined by visual inspection. From this we conclude that the coronary midplane should be estimated at least to within $(16 - 11.5)/2 = 2.25$ cm to ensure the full cardiac volume is contained within the reconstruction volume. A 1.6 cm error should be acceptable for this task. However, in subjects where femoral artery vascular access was used, the volume enclosed by the lowest plane containing the guide catheter (in the thoracic aorta) and the highest plane of the coronaries was 10 cm to 16.5 cm thick. Therefore, if both the coronaries and the guide catheter are to remain in focus simultaneously in the live image, a thicker reconstruction volume is likely necessary (e.g. $40 \times 5 \text{ mm} = 20 \text{ cm}$).

4. DISCUSSION

This work presents the first investigation of the automatic positioning of the tomosynthesis reconstruction volume of an inverse geometry fluoroscopy system during coronary angiography. Using human subject data, we demonstrate that sharpness metrics measured from a set of tomosynthesis planes of a non-contrast image frame can be used to predict the volume occupied by the coronary arteries. This technique could be applied prior to coronary angiography, or to assist in isocentering of the cardiac volume prior to rotational angiography.

Previous work has demonstrated that SBDX real-time tomosynthesis can be used to track multiple catheter elements in 3D with approximately 1 mm accuracy and precision.⁷ In real-time 3D catheter tracking, each discrete, high contrast element in the 2D display is localized along the source-detector axis by calculating the center-of-mass of the object's sharpness vs. tomosynthesis plane position. The cardiac localization algorithm presented here is a simplified form of this framework, where the depths of relatively large natural anatomic features (e.g. lung field) are localized along the source-detector axis to within ± 0.5 cm (the plane spacing). Despite this limitation, and the fact that a simple model established by linear regression was used to map feature locations to cardiac volume, prediction of cardiac volume position was sufficient for centering of the reconstruction volume.

Future work should evaluate whether improved accuracy can be obtained either through calculation of the center-of-mass of the sharpness curve, rather than peak position, or through the use of more sophisticated mapping models. Additionally, we note that the regional feature localization concept could be extended to track many smaller regions, potentially yielding 3D contours of the heart/lung interface or diaphragm/lung interface. Such an approach could be valuable in applications requiring cardiac gating or respiratory gating e.g. fusion of pre-acquired CT imaging with live SBDX imaging. To explore this possibility, we divided the image field-of-view into 10×10 pixel blocks and applied feature localization on each block. Each block was localized along the z-axis using the center-of-mass of the sharpness curve, and the x-y position was the center-of-mass within the 10×10 region at the peak plane. As shown in Figure 5, this preliminary investigation indicates that it may be feasible to localize the cardiomeastinal contour, diaphragm contour, and linear body of the catheter in 3D.

The cardiac localization algorithm uses a framework similar to real-time 3D catheter tracking and real-time composite reconstruction, both of which are implemented on the SBDX prototype. The current reconstructor has 8 GPUs dedicated to reconstructing tomosynthesis plane stacks, one GPU for composite image reconstruction, and one GPU for catheter tracking. It is expected that all image filtering and localization steps needed for cardiac localization could be handled by the catheter tracking GPU. The primary challenge is likely to be reconstruction of a wide plane stack (e.g. 88 planes) for a single scan frame. This task could be completed in 3 frame periods (0.2 s) if the plane positions were reprogrammed for each frame (32 planes/frame x 3 frames = 96 planes). Work underway to improve the computational efficiency of SBDX tomosynthesis may also enable reconstruction of a greater number of planes per frame, shortening the required time.

5. CONCLUSIONS

The position of the cardiac volume and the proper centering of the tomosynthesis reconstruction volume on an inverse geometry fluoroscopy system can be estimated from measurements of image sharpness versus plane position from a non-contrast image frame. Position estimate errors of 1–2 cm are acceptable for a 16 cm thick reconstruction volume. Regional measurements of sharpness enable 3D localization of soft tissue features.

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Figure 1.

The SBDX system (A) uses a scanning x-ray tube, multihole collimator, high speed detector, and real-time reconstructor (B). Each displayed fluoroscopic frame is a composite of a stack of tomosynthesis images (C) generated by unfiltered backprojection (shift-add). Each tomosynthesis image portrays in-plane and near-plane objects in focus.

Figure 2.

A: Example stack of 32 tomosynthesis images spaced by 5 mm, shown from lowest (top left) to highest (lower right) position relative to the x-ray tube. B: Composite image resulting from a properly centered tomosynthesis reconstruction volume. C: Effect of shifting the reconstruction volume 5 cm towards the x-ray tube, demonstrating loss of contrast and sharpness (yellow arrow) of an anterior coronary artery that falls outside the reconstruction volume. Contrast equalization has been applied to the images to facilitate visualization.

Figure 3.

Sharpness vs. plane position, calculated using the average gradient in the bright lung field (green curve), the number of normalized image gradients exceeding 0.5 (blue curve), and the maximum gradient in the image (red curve). Vertical dashed lines indicate peak locations in the sharpness curves. Example tomosynthesis images and gradient images corresponding to the sharpness peaks are shown at right. From top to bottom, there are peaks corresponding to sternal wires (51 cm), cardiomeastinal contour and diaphragm (44.5 cm), lung texture and guide catheter (38 cm), and the posterior ribs (33 cm) (see yellow arrows). The green outline on the upper right of the images encloses the bright lung field segmented by Otsu's thresholding method.¹⁴

Figure 4. Position of peak sharpness measured in a non-contrast frame of each angiographic loop. Sharpness metrics measured by average gradient in the bright lung field (green) and the number of gradients > 0.5 (blue) were correlated to the true coronary midplane.

Figure 5. MIP of gradient-filtered planes (A) and depth localization of 10x10 pixel regions (B,C). 3D spline curve fits (red) are shown for points localized along the cardiac contour, diaphragm, and catheter.