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Abstract. Scanning-beam digital x-ray (SBDX) is an inverse geometry x-ray fluoroscopy system capable of tomosynthesis-based 3-D catheter tracking. This work proposes a method of dose-reduced 3-D catheter tracking using dynamic electronic collimation (DEC) of the SBDX scanning x-ray tube. This is achieved through the selective deactivation of focal spot positions not needed for the catheter tracking task. The technique was retrospectively evaluated with SBDX detector data recorded during a phantom study. DEC imaging of a catheter tip at isocenter required 340 active focal spots per frame versus 4473 spots in full field-of-view (FOV) mode. The dose-area product (DAP) and peak skin dose (PSD) for DEC versus full FOV scanning were calculated using an SBDX Monte Carlo simulation code. The average DAP was reduced to 7.8% of the full FOV value, consistent with the relative number of active focal spots (7.6%). For image sequences with a moving catheter, PSD was 33.6% to 34.8% of the full FOV value. The root-mean-squared-deviation between DEC-based 3-D tracking coordinates and full FOV 3-D tracking coordinates was less than 0.1 mm. The 3-D distance between the tracked tip and the sheath centerline averaged 0.75 mm. DEC is a feasible method for dose reduction during SBDX 3-D catheter tracking. © 2017 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.4.2.023501](https://doi.org/10.1117/1.JMI.4.2.023501)]

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1 Introduction

Cardiac catheterization is a standard-of-care procedure for the diagnosis and treatment of structural heart disease and electrophysiological abnormalities.^{1–3} These procedures frequently involve the navigation of catheter devices within open 3-D spaces and the targeting of devices to specific anatomic positions. For example, radiofrequency catheter ablation of atrial fibrillation may require positioning of a catheter tip at points around the pulmonary vein ostia to create encircling lesions, which isolate irregular electrical signals. A structural heart intervention, such as transcatheter aortic valve replacement, requires accurate positioning of a stent-supported artificial valve within the aortic root prior to valve deployment. X-ray fluoroscopy provides essential real-time image guidance. However, in these procedures, the 2-D x-ray projection format does not convey 3-D device positions relative to target anatomy.

Three-dimensional tracking and display of catheter devices has the potential to reduce procedure duration and improve procedural outcome.^{4,5} Nonfluoroscopic 3-D catheter tracking systems, such as CartoMerge and NavX Fusion,⁴ are currently used in electrophysiology procedures. These systems rely on proprietary catheters fitted with electromagnetic field sensors and do not provide live imaging of the device relative to patient anatomy. Tracking can be merged with preacquired CT maps; however, if the reference catheter used for registration becomes displaced during the procedure, remapping and reregistration may be required.⁶ By contrast, tracking methods derived from fluoroscopy are capable of providing live anatomic context and

do not require modification of the device for tracking purposes, at the cost of increased radiation dose compared with nonfluoroscopic methods.^{7,8}

Scanning-beam digital x-ray (SBDX) is a low-dose, inverse-geometry x-ray fluoroscopy system with a unique real-time tomosynthesis capability.^{9–11} The system images the patient with an electronically scanned multisource x-ray tube, high speed photon counting detector, and real-time image reconstructor (Fig. 1).¹¹ In recent years, SBDX tomosynthesis has been explored for its potential for providing real-time 3-D tracking of catheter devices and fiducials simultaneous with fluoroscopy.^{12–14} A dose efficient design is achieved through inverse geometry beam scanning, which minimizes detected scatter and spreads primary x-rays over a relatively large skin entrance area.¹⁰ Recent work has shown that SBDX can achieve further dose reduction through fluence field modulation, i.e., selective downward-modulation of the outputs of individual focal spots in the multisource x-ray tube.¹⁵

When the x-ray imaging task is focused on a specific region of the field-of-view (FOV), an opportunity to reduce dose in the image periphery exists. For example, region-of-interest (ROI) fluoroscopy using a peripheral prepatient attenuator has been proposed for dose reduction during vascular interventions.¹⁶ Static, prepatient ROI attenuators have been shown to be capable of reducing dose-area-product (DAP) 1.5 to 3.5 times during imaging.¹⁷ Recent work has explored the use of moving attenuators in combination with 2-D image-based object tracking to reduce the dose in regions around a moving stent.¹⁸ Currently, two commercial solutions (ControlRad, Inc. and

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Fig. 1 (a) SBDX prototype. (b) An electron beam scans over an array of focal spot positions. A multihole collimator defines overlapping x-ray beamlets directed at the detector. (c) A fixed point in the imaging volume is imaged by a patch of focal spots. (d) Multiple tomosynthesis planes are formed by back-projecting detector images onto planes of reconstruction. Object sharpness versus plane position is used for 3-D localization.

IKOMED Technologies Inc.) exist; they use mechanical ROI attenuators to define a moving ROI and have demonstrated significant dose reduction.^{19,20} For tasks that do not require any peripheral information, as in SBDX 3-D catheter tracking, moving collimation could be employed.

This work proposes a new dose-reduced technique for SBDX catheter tracking, which combines tomosynthesis-based device localization with dynamic electronic collimation (DEC) of the scan. Electronic collimation refers to blanking of the tube current at selected focal spot positions to reduce the x-ray field area. It is hypothesized that frame-by-frame electronic collimation to a region just surrounding the tracked catheter will enable a major additional reduction of DAP without loss of tracking accuracy. Furthermore, this technique may reduce peak skin dose (PSD) when the device is moving across the FOV. The performance and feasibility of the new technique was evaluated in phantom and Monte Carlo studies, which compared DEC-based tracking with full FOV SBDX tracking.

1.1 Scanning-Beam Digital X-Ray

SBDX uses a multisource x-ray tube to scan a narrow x-ray beam in an inverse geometry (Fig. 1). This geometry is a form of real-time tomosynthesis. An electron beam is raster scanned over a 2-D array of focal spot positions on a large area transmission style tungsten target (up to 100×100 positions, 2.3-mm pitch). A multihole collimator between the target and patient defines a series of overlapping x-ray beamlets, each of which is directed at a high speed photon counting detector array. As the electron beam moves from one hole to the next, detector images are simultaneously captured. The current prototype has a 5.3 cm $\times$ 10.6 cm detector (0.33-mm native element pitch) positioned 150 cm above the target plane. Mechanical isocenter is 45 cm above the target. In a typical 15 frame/s imaging mode, a 71×71 source array is used and the electron beam visits each focal spot position eight times per frame in a blockwise raster pattern.^{9,11} The stream of detector images captured during scanning is reconstructed into a stack of tomosynthesis images in real time using unfiltered (shift-and-add) back projection.^{11,21}

The current GPU-based reconstructor generates 32 planes at 15 frame/s with programmable plane-to-plane spacing. A composite image of the plane stack is formed for real-time display by selecting in-focus features from each plane and displaying them in a single image. Additional details of SBDX operating principles can be found in Refs. 9 and 11.

The SBDX skin dose reduction strategy is based on: (i) narrow beam scanning with a large air gap, which minimizes detected x-ray scatter, (ii) efficient detection of primary x-rays using a thick CdTe detector, and (iii) an inverse geometry, which spreads entrance x-rays over a large area. A study with a previous SBDX prototype found that these properties increase entrance exposure efficiency by three to seven times when compared with a conventional fluoroscopic system.¹⁰ Further dose reduction can be achieved by tailoring the outputs of individual focal spots according to patient geometry and imaging application. Specifically, the SBDX x-ray tube has a grid-controlled tube current, which can be turned on or off at each focal spot position, allowing for nonmechanical 2-D collimation and/or spatial modulation of the x-ray field (e.g., by turning of the beam for a fraction of the eight passes per frame). A technique termed regional adaptive exposure (RAE) scanning performs spatial modulation of the x-ray tube output. By reducing the number of focal spot activations per frame at source positions with high measured x-ray transmission, this method has been shown to reduce DAP by an additional 46%.¹⁵ In this paper, we combine the 3-D localization capabilities of SBDX with electronic collimation in the x-ray tube to minimize the catheter tracking radiation dose.

1.2 Tomosynthesis-Based 3-D Catheter Tracking

The stack of SBDX tomosynthesis images reconstructed in a frame period can be analyzed to localize discrete, high-contrast catheter elements in 3-D with submillimeter accuracy and precision.^{12–14} The SBDX tracking algorithm is based on the analysis of tomographic blurring versus plane position, as outlined in Fig. 2. For each image in the plane stack, a low-pass filter is applied to suppress image noise. Next, a gradient filter

Fig. 2 Flowchart of SBDX 3-D catheter tracking. (a) The tomosynthesis image stack reconstructed for a frame period is the input to the tracking algorithm. (b) Gradient filters are applied to produce a “score stack” reflecting the contrast and sharpness at each pixel in each plane. (c) A MIP along the z -axis is created from the score stack. (d) A threshold is applied, and connected component labeling is used to find distinct objects. (e) A weighted center-of-mass calculation is performed on each object for x - y localization. (f) Score versus plane position is extracted for each object. (g) The z -coordinate of an object is the center-of-mass of the distribution of scores versus z -plane, after application of a second threshold to reject background scores.

(Prewitt kernel) is applied to emphasize in-focus and high-contrast features. A second low-pass filter is applied to distribute high gradient edges over the object area and aid in subsequent segmentation of objects. The size of this kernel is adjusted to the size of the object being tracked. For efficient object segmentation, a maximum intensity projection (MIP) of the gradient filtered stack is formed along the source-detector direction (z -axis). A threshold is applied to the MIP, and the surviving “blobs” (objects) are labeled using connected component labeling. A weighted center-of-mass is calculated for each labeled object to localize the object position in the x - y plane. To localize the object position along the source-detector (z) axis, the object’s gradient values are extracted from each z -plane. Since SBDX tomographic blur is symmetrical about the object position, the measurements of sharpness versus plane position are treated as a probability distribution function of object position. The z -coordinate is determined by calculating the center-of-mass of this distribution, after application of a second threshold, which rejects background gradients.

2 Methods

2.1 Dynamic Electronic Collimation

In the proposed DEC method, a small ROI of focal spots is turned on in each frame to create an x-ray field that just surrounds the tracked object. Inside the ROI, all passes of the electron beam are performed with normal tube current. For example, in the $71 \times 71 \times 15$ frame/s scanning mode, which involves eight passes per hole per frame period, all eight passes would be performed at normal tube current inside the ROI. Outside the ROI, the tube current is turned off for all passes. The position of the ROI is updated after each imaging frame based on previous tracking results. The reasoning behind the DEC algorithm is twofold. First, accurate SBDX tracking requires only the image

information that captures the local tomographic blurring of the object. Second, it is assumed that once the operator has switched to a 3-D tracking mode, the patient anatomy of interest lies near the device being tracked. We note that the DEC technique could also be modified to create a “soft ROI,” where the tube current is left on for a small number of passes (e.g., one out of eight passes) at all focal spot positions. Such a mode could be employed if peripheral image context is desired.

To prevent a reduction in catheter tracking accuracy and precision, the ROI must be defined, so the number and angular range of rays passing through the object are not reduced relative to full FOV mode. For the unfiltered back projection reconstruction method used with SBDX, a point in a full FOV image receives fluence contributions only from the x-ray beamlets, which overlap that point. Therefore, the minimum allowed ROI width for a point-like object can be estimated by tracing the edges of the detector back through the object and onto the source plane [see Fig. 1(c)]:

$$n_x = \frac{W_x}{\Delta_x} \left(\frac{\text{SDD}}{z} - 1 \right)^{-1} \quad (\text{width in \# focal spots}), \quad (1)$$

where W_x is the detector width (106 mm) measured along the direction of the scan lines, Δ_x is the focal spot pitch along the scan line direction (2.3 mm), z is the source-to-object distance, and SDD is the source-to-detector distance (1500 mm).¹¹ A similar expression exists for the ROI height:

$$n_y = \frac{W_y}{\Delta_y} \left(\frac{\text{SDD}}{z} - 1 \right)^{-1} \quad (\text{height in \# focal spots}), \quad (2)$$

where W_y is the detector height (53 mm) and Δ_y is the focal spot pitch perpendicular to the scan lines (2.3 mm). For an object with substantial width, additional focal spot positions may

Fig. 3 Calculation of the ROI width for an object located at distance z from the source plane. A single point in the image volume is imaged by n_x focal spot positions. For an object of width d_x , an additional n_x^{add} source points should be used.

need to be added to the ROI. As shown in Fig. 3, if the object has in-plane dimensions d_x and d_y in the x and y directions, respectively; then the number of focal spots in the ROI should be extended by an amount

$$n_{x,y}^{add} = \frac{d_{x,y}}{\Delta_{x,y}} \left(1 - \frac{z}{\text{SDD}} \right)^{-1} \quad (3)$$

(# extra focal spots, in x or y directions)

in the x and y directions.

The ROI size determined by Eqs. (1)–(3) depends on object position z , which in theory could be obtained from catheter tracking. In this paper, we investigate the simpler case of a moving fixed-size ROI. For a fixed-size ROI, a conservative estimate of ROI size is obtained by assuming that the object is at the highest plane of the reconstructed volume ($z = 525$ mm). For a $10.6 \text{ cm} \times 5.3 \text{ cm}$ detector, this approach gives a ROI size of 24×12 active sources. To account for the finite size of a catheter tip, two additional source points were added in each direction, giving a 26×14 ROI. Last, we note that the corner focal spot positions in a rectangular ROI make a minimal contribution to the fluence at the center of the reconstructed ROI. This is because the x-ray beamlet itself is not perfectly rectangular in shape and tapers at the corners. Based on this observation, we used an elliptical ROI with 26 sources along the major axis and 14 sources along the minor axis.

Figure 4 shows an example DEC scan map and image reconstruction for an elliptical 26×14 ROI with a total of 340 active focal spots. In comparison, the full FOV image uses 4473 active focal spots (71×71 array with rounded corners). The relative DAP for DEC imaging versus full FOV imaging should be approximately equal to the ratio of active focal spots in the two cases ($340/4473 = 0.076$). The reduction of PSD is dependent on the object trajectory in an image sequence. In an individual frame, the PSD in DEC mode is expected to be similar to the full FOV PSD. However, if the object is moving in the image sequence and the ROI is moving across the skin entrance, then the total PSD is determined by the total x-ray field overlap time at a skin point, rather than the total imaging time. Therefore, significant PSD reduction can be expected for wide-ranging catheter trajectories.

When using the DEC algorithm, the initial 3-D position of the catheter is not known *a priori*. Furthermore, the scan map must be updated for each frame to reflect the predicted catheter position. Therefore, the following algorithm is employed to initialize and update the DEC scan (see Fig. 5). Frames 1 and 2 are acquired with full FOV scanning. Objects of interest are localized in these frames, and the localized coordinates from frames 1 and 2 are used to calculate a velocity vector for tracked objects. The velocity vector is then used to predict the position of the object in frame 3, and an ROI scan map is generated for this predicted position. Using this scan map, the third frame is acquired with a reduced FOV, and object localization is again performed. The velocity vector is updated using object positions from frames 2 and 3, and the ROI position is updated for frame 4. The procedure is repeated for the rest of the imaging sequence. By this method, the catheter positions in the two

Fig. 4 (a) Full FOV image acquisition. (b) DEC-mode scan map generated from localized catheter tip, indicating positions of active focal spots. (c) Image corresponding to the DEC-mode scan.

Fig. 5 DEC imaging method. To initialize, full FOV imaging is performed for two frames. 3-D tip localization is performed for both frames, and a velocity vector (blue) is calculated. The velocity vector is used to predict the position of the tip in frame 3. The source is electronically collimated around this predicted position (third panel), and frame 3 is acquired in DEC mode. Subsequently, the velocity vector and DEC scan map are continuously updated using the two most recent frames of catheter tracking. (a) Frame 1, (b) frame 2, (c) frame 3 ROI, (d) frame 3, and (e) frame 4 ROI.

Algorithm 1: Initialize DEC

```

1: For frames:  $f_{i-1}$  and  $f_i$ 
2:   Acquire full field-of-view images
3:   Localize object positions:  $p_{i-1}$  &  $p_i$ 
4:   Calculate velocity vector:  $v_i = p_i - p_{i-1}$ 
5: End for

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Algorithm 2: DEC Imaging

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1: If velocity vector exists for  $f_i$ 
2:   ROI placement for  $f_{i+1}$  at estimated position:  $p_{i+1}^* = p_i + v_i$ 
3:   Acquire  $f_{i+1}$  ROI image
4:   Localize true object position:  $p_{i+1}$ 
5:   If no points are localized
6:     Initialize DEC
7:   Else
8:     Update velocity vector:  $v_{i+1} = p_{i+1} - p_i$ 
9:   End if
10: End if

```

Fig. 6 DEC algorithm.

most recent frames are used to generate a velocity vector that defines the ROI position for the next frame. The algorithm is summarized in Fig. 6. If no catheter is present or the catheter is not detected, full FOV images are acquired and displayed until localization is achieved for two consecutive frames, after which DEC imaging begins. Images are always displayed, regardless of the outcome of the catheter tracking algorithm.

2.2 Tracking Performance

DEC tracking versus full FOV tracking was evaluated in a phantom study. A cardiac chamber phantom (Venous Sam, Venous Training Model 5010, Lake Forest Anatomicals, Lake Forest, Illinois) was fixed within an anthropomorphic chest phantom. A catheter sheath was mounted within the phantom to define a trajectory from the inferior vena cava into the right atrium, through the atrial septum, and into the left atrium. The tracked position of a catheter tip pulled through this sheath was compared with the volume and centerline of the sheath, as defined by CT, following the method described in Ref. 13. The experimental setup and orthogonal projections showing the chest wall and sheath are shown in Fig. 7. The phantom contained fiducials fixed to the chest wall for registration of SBDX and CT data.

The phantom was imaged in the 71×71 , 15 frame/s cardiac scan mode as a 7 Fr ablation catheter with 4 mm tip (Livewire TC, St. Jude Medical, Inc.) was pulled along the 3-D trajectory defined by the sheath by a linear motion stage. Image sequences

were recorded for pullback speeds of 10, 25, and 50 mm/s. Raw detector data were saved for each collimator hole for retrospective generation of DEC scans. This retrospective approach enabled evaluation of tracking performance for different DEC scan modes using the same underlying phantom geometry and motion. The DEC algorithm described in Sec. 2.1 was implemented offline, using MATLAB versions of the SBDX image reconstruction and catheter tracking algorithms. The DEC images were reconstructed with the detector data originating from within the 26×14 elliptical ROI, centered at the position predicted by catheter tracking. Full FOV reconstruction and catheter tracking were performed for comparison. Full FOV images were reconstructed using data from all collimator holes.

For both DEC and full FOV imaging, tomosynthesis reconstruction was performed for 32 planes with 5-mm spacing centered on the mechanical isocenter. The catheter tracking parameters were adjusted to localize the catheter tip while rejecting the fiducials in the image.^{12,13} The initial boxcar filter width was 6 pixels, the secondary boxcar width was 14 pixels, and the XY threshold and z threshold were each set to 22%.

The root-mean-squared-deviation (RMSD) between the DEC-based catheter tip positions (p_i) and the full FOV-based catheter tip positions was calculated to assess any change in localization resulting from DEC scanning:

$$\text{RMSD} = \sqrt{\frac{1}{n} \sum_{i=1}^n \| \mathbf{p}_i^{\text{DEC}} - \mathbf{p}_i^{\text{Full-FOV}} \|^2}. \quad (4)$$

Additionally, the 3-D tip positions were compared with the 3-D volume and centerline of the catheter sheath. To define the sheath volume, a helical CT acquisition of the anthropomorphic phantom was performed on a GE 750 HD scanner at 120 kVp, 295 mAs. CT reconstruction was performed with 0.625-mm slice thickness and $0.717 \text{ mm} \times 0.717 \text{ mm}$ in-plane pixel size. The sheath volume was segmented using intensity-based segmentation and a centerline was extracted from the sheath volume. A 3-D–3-D transformation from SBDX coordinates to CT coordinates was calculated with an iterative closest point rigid registration algorithm, using the 3-D fiducial positions localized from a 72-plane SBDX reconstruction and the 3-D fiducial positions segmented from the CT scan. This transformation was applied to the localized catheter tip coordinates for each frame to transform the tip position from SBDX coordinates to CT coordinates. The accuracy of tip tracking was characterized in terms of the fraction of the localized tip positions falling within the sheath volume, the mean distance from the localized tip to the sheath centerline, and the standard deviation in tip-to-centerline distances.

2.3 Dose Reduction Study

A study was performed to determine the dose reduction characteristics of DEC imaging. Currently, the SBDX prototype has a static electronic collimation feature, which uses a stationary circular ROI of focal spots with a specified position and radius.¹⁴ This feature was used to measure DAP versus ROI size. The dose reduction associated with a moving elliptical ROI was evaluated using a Monte Carlo simulation package developed for the SBDX geometry.²² This simulation package has been previously validated against relative DAP measurements performed in RAE versus full FOV scanning mode. To confirm that this package is also valid for simulated ROI scanning,

Fig. 7 (a) Experimental setup showing an anthropomorphic chest phantom, fixed catheter sheath entering a vascular phantom, and catheter attached to a linear motion stage. (b–d) Orthogonal projections of a CT scan of the phantom, showing the 3-D path of the internal sheath.

the DAP measurements made with static circular ROIs were compared against Monte Carlo simulations of the same ROI scanning modes.

In the experimental study of circular scan ROIs, electronic collimation was performed with ROI radii of 3, 5, 10, 15, 25, and 35 collimator holes. Each ROI was centered on the multihole collimator. DAP was measured with a 300 mm × 300 mm active area DAP meter (Radcal DAPcheck PDC, Radcal Corporation, Monrovia, California) placed just above the collimator exit (6 cm from target plane). For each collimation radius, five 20-s sequences were acquired. The x-ray tube was operated at 100-kVp and 120-mA peak tube current. Five, 20-s sequences for a full FOV acquisition were also performed for use as a reference. From these acquisitions, the DAP rate was calculated in units of $\mu\text{Gy} \cdot \text{m}^2 / \text{min}$.

A Monte Carlo package based on MC-GPU v1.3 and adapted for inverse geometry fluoroscopy²² was used to simulate a moving elliptical ROI in DEC scanning. MC-GPU is a public domain Monte Carlo x-ray transport package based on the PENELOPE subroutine package.^{23–25} This code has been modified to simulate the SBDX imaging geometry. User inputs include a voxelized phantom geometry, source x-ray energy spectrum, and scan maps, which specify the relative x-ray output from each focal spot position. The voxelized kerma output tally was used to estimate dose. To confirm the ability of this package to simulate an ROI scan mode, simulated scan maps were generated to match those used in the experimental DAP study. The simulated phantom geometry was a volume of air measuring 20.3 cm × 20.3 cm × 70 cm with 1.59-mm isotropic voxels. The DAP was calculated from simulation by integration of the 2-D kerma distribution $K(i, j)$ at the plane corresponding to the position of the DAP meter:

$$\text{DAP}_{\text{MC}} = \sum_{i=1}^m \sum_{j=1}^n K(i, j) \Delta x \Delta y, \quad (5)$$

where $\Delta x \Delta y$ is the in-plane voxel area and the plane contains $m \times n$ voxels. For comparison with experimental data, Monte Carlo DAP measurements versus ROI size were multiplied by a calibration factor equal to the ratio of the measured full FOV DAP divided by the simulated full FOV DAP.

Following the validation study, Monte Carlo simulation was used to estimate the DEC and full FOV doses corresponding to the tracking study (Sec. 2.2). The simulated patient was a voxelized anthropomorphic phantom derived from a CT of an adult chest phantom (CIRS Adult Male Atom Dosimetry Verification Phantom model 701, CIRS Inc., Norfolk, Virginia) placed in a head-first, supine position with the entrance plane to the phantom positioned 15 cm below the mechanical isocenter. The voxel dimensions were 0.781 mm × 0.781 mm × 0.625 mm. DAP, integral dose, and PSD were calculated from the voxelized kerma output of each simulation. DAP was estimated by integrating kerma in the plane of air just posterior to the simulated patient's back. PSD was defined as the maximum kerma in the same plane, after convolution with a 2-D Gaussian kernel ($\sigma = 2.3$ mm) for noise reduction. Integral dose was determined by integrating the kerma distribution over the entire phantom volume. These dosimetric quantities were determined for both a single frame of imaging and for the entire pullback imaging sequence.

3 Results

3.1 Selection of Region-of-Interest Size

The scan ROI dimensions needed to fully illuminate a tracked object are calculated from system geometry, object position, and object size. As discussed in Sec. 2.1, a 24×12 elliptical ROI was expected to be adequate for a pointlike object, and a 26×14 elliptical ROI was calculated for the finite-sized catheter tip used in this study (7 Fr × 4 mm). To check that tracking precision would be similar to full FOV precision with this ROI size, DEC-based catheter tip tracking was evaluated for ROI sizes

Fig. 8 RMSD between of catheter positions localized from DEC versus full FOV imaging, as a function of the ROI size. Results are shown for elliptical ROIs with 2:1 major-to-minor axis ratio.

ranging from 8×4 to 44×22 . The catheter tip was imaged at the mechanical isocenter with a mean isocenter-plane fluence of 450 photons per reconstructed pixel per frame under full FOV conditions (0.161 mm pixel width). The RMSD between the DEC-based catheter tip positions and the full FOV-based positions was calculated from 25 frames of imaging. Figure 8 presents RMSD as a function of the minor axis of the elliptical ROI. The RMSD was 0.03 mm when the ROI was a 26×14 ellipse. Smaller scan ROIs yielded less precise tracking, as expected from the effects of reduced fluence and reduced effective tomographic angle. Using the 24×12 ROI predicted for a point-like object, the RMSD was 0.20 mm.

3.2 Tracking Performance

Image sequences were acquired for 10, 25, and 50 mm/s catheter pullback speeds. The image intensity was 499 ± 50 photons per reconstructed pixel at the sheath volume in the lung field and 290 ± 30 photons/pixel in the spine region. Figure 9 shows an example of SBDX 3-D catheter tip tracking relative to the sheath volume. Tip positions found within the intensity gradient at the edge of the image (12% of the image width laterally and 6% vertically) were omitted from analysis since accurate localization at the image edges is known to be challenging. The tracking analysis results are listed in Table 1. The RMSD

Fig. 9 SBDX tracking of catheter tip (red points), catheter sheath volume (light gray), and sheath centerline (black) for the 10-mm/s pullback sequence. The tip-to-centerline distances are shown with blue lines.

between the DEC-based and full FOV tip positions was 0.076, 0.064, and 0.075 mm for the 10-, 25-, and 50-mm/s pullback sequences, respectively. For reference, the RMSD values were less than the dimension of one image pixel (0.161 mm at isocenter).

In the full FOV sequences, the mean 3-D distance from the localized catheter tip to the sheath centerline was 0.75, 0.65, and 0.63 mm for 10-, 25-, and 50-mm/s pullback speeds, respectively. In the DEC sequences, the mean tip-to-centerline distances were 0.77, 0.79, and 0.68 mm for 10-, 25-, and 50-mm/s pullback speeds, respectively. For both scanning modes, 100% of the analyzed tip positions were contained within the sheath volume, and the standard deviation in tip-to-centerline distance averaged 0.39 mm. Note that deviations between the tracked tip and the sheath centerline may be due to actual deviations of the catheter tip from the sheath centerline, stochastic variation in tip-tracking (± 0.2 mm), and registration error between SBDX and CT coordinate systems (0.14 mm). The target registration error of the rigid SBDX-to-CT transformation was calculated by the methods in Ref. 26, using the root-mean-square fiducial localization error, the fiducial positions, and the points along the sheath centerline as target positions.

3.3 Dose Reduction Study

Measurements of DAP rate versus circular scan ROI size for 100-kV, 120-mA peak tube operation are presented in Fig. 10. The DAP rate ranged from $5.32 \mu\text{Gy} \cdot \text{m}^2/\text{min}$ using a 3 spot radius (29 active spots) to $916 \mu\text{Gy} \cdot \text{m}^2/\text{min}$ using a 35 spot radius (3835 active spots). As expected, the measured DAP scaled linearly with the number of active focal spots in the scan ($Y = 0.2393 \cdot X$; $R^2 = 0.988$). The Monte Carlo simulations of DAP versus ROI size also demonstrated a linear dependence on the number of active spots. From the simulation, it was calculated that a DEC scan with a 340-spot elliptical scan ROI would produce a DAP rate equal to 7.8% of the full FOV DAP rate (4473 spots), which measured $1050 \mu\text{Gy} \cdot \text{m}^2/\text{min}$ in this experiment.

Monte Carlo simulations of air kerma at the phantom entrance plane for a single frame of DEC and full FOV imaging are shown in Figs. 11(a) and 11(c), respectively. Axial slices of the corresponding kerma distributions are shown in Figs. 11(b) and 11(d). The single-frame DEC DAP was 8.4% of the full FOV DAP, and the integral kerma was 8.1% of the full FOV value. DAP and integral dose reductions were consistent with the relative number of active focal spots used in DEC versus full FOV scanning (7.6%). PSD from the single-frame DEC simulation was 84.8% ($\pm 0.8\%$) of the full FOV value. It was expected that the peak primary x-ray fluence in a single frame of imaging would be essentially unchanged for DEC mode, and, consistent with this expectation, when the simulated phantom was replaced with air, the single-frame DEC PSD was 99.6% of the full FOV value. The slight reduction in PSD observed with the phantom in place is consistent with a greater backscatter contribution to the entrance plane when using full FOV mode.

In simulations of a full “image sequence” of DEC catheter tracking, the DAP reduction was similar to the single frame case and the PSD reduction was greater than the single-frame case. Air kerma distributions at the phantom entrance plane for the 50-mm/s pullback image sequence are shown in Fig. 12. Figure 13 shows an outline of the image ROIs corresponding to each frame of the DEC scan. The DEC-mode DAP was

Table 1 Comparison of catheter tracking performance and dose for DEC versus full FOV imaging.

Sequence	Number of frames #	DEC versus full FOV tracking	Average tracked tip-to-centerline distance		Dose ratio, (DEC/full FOV) ^a		
		RMSD (mm)	DEC (mm)	full FOV (mm)	DAP %	PSD %	integral dose %
10 mm/s pullback	154	0.076	0.77	0.75	7.4	33.6	7.2
25 mm/s pullback	67	0.064	0.79	0.65	7.8	34.8	7.7
50 mm/s pullback	38	0.075	0.68	0.63	7.6	34.0	7.5
Single frame	1	—	—	—	8.4	84.8	8.4
Single frame, no phantom	1	—	—	—	7.9	99.6	7.9

^aMonte Carlo based percentages have ±0.8% estimated uncertainty.

Fig. 10 Measured DAP rate versus the number of active focal spots. The static electronic collimation feature on SBDX was used to create different scan ROIs. The highest value corresponds to a full FOV. The Monte Carlo simulated DAP rates for the same ROIs are shown for comparison (full FOV case excluded).

Fig. 11 Air kerma distributions for a single frame of DEC and full FOV imaging. (a) Entrance-plane air kerma for DEC imaging, (b) axial slice of kerma for DEC imaging, (c and d) entrance plane and axial slice of kerma for full FOV imaging.

Fig. 12 Air kerma at entrance plane for a full image sequence (50 mm/s pullback) using (a) DEC scanning and (b) full FOV imaging.

7.4%, 7.8%, and 7.6% of the full FOV DAP for the 10, 25, and 50 mm/s sequences, respectively. DAP reduction was consistent with the reduction in active focal spots. The DEC-mode PSD was 33.6%, 34.8%, and 34.0% of the full FOV PSD for the three sequences. In each sequence, the PSD was located at the top of the catheter trajectory, where motion was parallel to the long axis of the elliptical scan ROI. The additional PSD reduction observed for the full image sequence versus the single-frame case is due to the fact that, in DEC mode, the dose scales with the number of x-ray fields overlapping a given point, whereas in full FOV mode, the dose scales with the number of image frames. When the catheter and scan ROI are moving, the number of overlapping x-ray fields is reduced and a PSD reduction is achieved.

Reduction of PSD in the DEC image sequence is further demonstrated in Fig. 13. Air kerma rate and cumulative air kerma were monitored in a plane 15 cm below isocenter, directly below a point on the catheter path (marked with a star in Fig. 13). The primary x-ray kerma rates for DEC and full FOV scanning were nearly identical when the DEC x-ray field was overlapping the selected point. However, this point was irradiated for a shorter period of time and for fewer frames in DEC mode, leading to a lower cumulative kerma at the end of the image sequence.

4 Discussion

This study reports a DEC method designed to minimize the total dose associated with SBDX 3-D catheter tracking. This first

Fig. 13 (a) Full FOV image and ROIs (blue) indicating path of the collimated x-ray field during DEC-mode catheter tracking. (b and c) Cumulative air kerma and air kerma rate from primary x-rays at a point along the catheter path [star on (a)] in the entrance plane. In full FOV mode, the point is irradiated for the full image sequence. In DEC mode, the point is briefly irradiated as the collimated x-ray field passes by.

investigation of DEC imaging indicates that the DAP may be reduced by an order of magnitude when tracking a single object, with negligible changes in tracking accuracy and precision. Reduction in PSD is also possible. These reductions are in addition to the existing dose efficiency of the SBDX system.

In this study, the DAP was reduced to 8% of the full FOV value on average. PSD was 34% of the full FOV value for image sequences with a moving catheter and 85% for a single frame of imaging. The degree of PSD reduction over an image sequence is dependent on the catheter trajectory and device size. A single object was tracked along a relatively simple trajectory in this phantom study. Further studies are needed to determine PSD reduction for more realistic catheter motions encountered in clinical procedures. These studies should also investigate DAP and PSD reduction in scenarios where multiple catheter devices of interest are present. In this scenario, the DEC mode scan for an image frame would be the union of the ROIs calculated for the individually tracked catheter electrodes. Specific catheters can also be identified from SBDX tracking results based on the known configuration of the catheter electrodes, as was recently demonstrated for coronary sinus catheter detection.²⁷

The DEC algorithm requires accurate tracking and prediction of catheter positions, so the active scan region can be continuously updated to remain centered on the catheter. Catheter positions were accurately predicted in this initial study. However, in clinical procedures, more complicated user-induced and cardio-respiratory motion can be expected. Two types of challenges are anticipated. First, the tracked object could fall outside the SBDX FOV due to respiratory motion. In this event, the DEC algorithm would reinitialize with full FOV imaging. This problem can be minimized through attention to patient centering. For example, the 11.4-cm FOV can fully enclose the left atrium (4.03 ± 0.77 cm),²⁸ and an ablation catheter inside a centered atrium should remain in the FOV even with displacements caused by respiratory motion. Second, reversals in catheter motion could lead to errors in motion prediction and ROI centering. The ROI size could be increased beyond the minimum size to accommodate errors in motion prediction, potentially

improving robustness of the algorithm while still providing dose reduction relative to full FOV imaging. Similarly, the ROIs from several previous frames could be allowed to persist to allow for reversal of motion. Alternatively, the number of incorrect predictions of catheter position could be reduced by periodically performing a full FOV scan frame or using a “soft ROI,” which maintains low-intensity illumination of the region outside the ROI, similar to ROI fluoroscopy.¹⁶ Since the electron beam visits each collimator hole multiple times in an image frame, a soft ROI could be implemented by blanking the electron beam for a fraction of the beam visits to the peripheral holes.

Further dose optimization may be possible for the DEC algorithm. In particular, this study did not consider the tradeoff between image SNR inside the ROI and tracking accuracy. It is possible that the desired tracking performance in a clinical procedure can be achieved with lower dose levels than were examined in this study. It may also be possible to adaptively reduce the x-ray output from the active focal spots as the ROI moves from lower to higher transmission regions of the patient (e.g., from mediastinum to lung). Conceptually, this would be similar to the RAE control method, which has been reported for full FOV SBDX imaging.¹⁵

Although a real-time implementation of DEC was beyond the scope of the present study, the algorithm presented here is suitable for real-time imaging. The algorithm for 3-D catheter tracking from real-time SBDX tomosynthesis was recently implemented in the SBDX imaging chain.²⁹ When imaging at 15 frame/s, 3-D tracking coordinates can be computed in 41.5 ms, leaving ~ 25 ms/frame for additional computations. Calculation of velocity vectors from two frames worth of tracking results, as needed for DEC, can be accomplished in less than 1 ms in our current MATLAB implementation.

5 Conclusions

DEC is a feasible method for reducing dose and DAP during SBDX catheter tip tracking. For imaging an ablation catheter tip, DAP may be reduced to 8% of full FOV levels with negligible change in SBDX catheter tracking performance. PSD from

an imaging series can also be reduced for moving devices. This technique is expected to be an important component of a dose-optimized 3-D tracking method for SBDX. Further investigation in clinically realistic imaging scenarios is needed to characterize and optimize the algorithm.

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