Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2015 ; 9412: . doi:10.1117/12.2081716.

Detector, collimator and real-time reconstructor for a new scanning-beam digital x-ray (SBDX) prototype

Michael A. Speidel^{a,b}, Michael T. Tomkowiak^a, Amish N. Raval^b, David A. P. Dunkerley^a, Jordan M. Slagowski^a, Paul Kahn^c, Jamie Ku^c, and Tobias Funk^c

^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

Scanning-beam digital x-ray (SBDX) is an inverse geometry fluoroscopy system for low dose cardiac imaging. The use of a narrow scanned x-ray beam in SBDX reduces detected x-ray scatter and improves dose efficiency, however the tight beam collimation also limits the maximum achievable x-ray fluence. To increase the fluence available for imaging, we have constructed a new SBDX prototype with a wider x-ray beam, larger-area detector, and new real-time image reconstructor. Imaging is performed with a scanning source that generates 40,328 narrow overlapping projections from 71×71 focal spot positions for every 1/15 s scan period. A high speed 2-mm thick CdTe photon counting detector was constructed with 320×160 elements and $10.6 \text{ cm} \times 5.3 \text{ cm}$ area (full readout every $1.28 \mu\text{s}$), providing an 86% increase in area over the previous SBDX prototype. A matching multihole collimator was fabricated from layers of tungsten, brass, and lead, and a multi-GPU reconstructor was assembled to reconstruct the stream of captured detector images into full field-of-view images in real time. Thirty-two tomosynthetic planes spaced by 5 mm plus a multiplane composite image are produced for each scan frame. Noise equivalent quanta on the new SBDX prototype measured 63%–71% higher than the previous prototype. X-ray scatter fraction was 3.9–7.8% when imaging 23.3–32.6 cm acrylic phantoms, versus 2.3–4.2% with the previous prototype. Coronary angiographic imaging at 15 frame/s was successfully performed on the new SBDX prototype, with live display of either a multiplane composite or single plane image.

Keywords

x-ray fluoroscopy; scanning-beam digital x-ray; photon-counting detector; real-time tomosynthesis

1. INTRODUCTION

Scanning-beam digital x-ray (SBDX) is an inverse geometry x-ray fluoroscopic technology designed for dose reduction and 3D device guidance in cardiac interventional procedures (Figure 1).¹ The dose reduction strategy is based upon i) reducing image-degrading scatter through the use of a narrow x-ray beam and large air gap, ii) imaging in an inverse geometry that distributes x-rays over a larger area at the skin entrance, and iii) utilizing a thick CdTe

detector that maintains high efficiency at high tube voltage.² The inverted scanning geometry also gives SBDX a unique real-time tomosynthesis capability which may be exploited to extract 3D information about anatomy or devices. Potential applications of this capability include lung tomosynthesis,³ frame-by-frame 3D catheter tracking,⁴ stereoscopic fluoroscopy,⁵ and calibration-free vessel sizing.⁶

A previously reported SBDX prototype demonstrated potential for a several-fold reduction in entrance dose,² however achieving cine-quality imaging was challenging due to the limitation on maximum x-ray output and image fluence imposed by tight beam collimation. In this work we describe the construction of a new SBDX prototype designed to reduce this limitation through use of a wider x-ray beam, larger x-ray detector, and new real-time image reconstruction hardware. Improvements to detected photon fluence and example coronary angiographic images in a human subject are presented.

2. BACKGROUND

2.1. Inverse geometry x-ray fluoroscopy

Figure 1 shows the new SBDX system (Triple Ring Technologies, Inc; NovaRay Medical Inc., Newark, CA) and the principles of inverse geometry fluoroscopy. The system consists of a large-area x-ray tube with a 2D array of focal spot positions, a multihole collimator, a high speed photon counting detector, and a real-time image reconstructor. For each fluoroscopic image frame, a series of narrow overlapping x-ray projections is captured and reconstructed into a full field-of-view image. The x-ray tube and detector are mounted to a rotating C-arm with 45 cm source-to-isocenter distance and 150 cm source-to-detector distance (SDD).

Inside the x-ray tube an electron beam is electromagnetically deflected over a thin transmission-style tungsten target, dwelling behind each collimator hole for 1.04 μ s and moving between holes in 0.24 μ s. As scanning proceeds, detector images are captured and streamed to the reconstructor. Several scanning modes are possible, providing different fields-of-view and frame rates.¹ The full collimator has a ‘squarcle’ shape with 100 holes across and 100 holes from top to bottom. Reduced field-of-view imaging is achieved by scanning a subset of the holes. In the 71 $\times$ 71 15 frame/s scan mode, 71 $\times$ 71 focal spot positions on a 2.3 mm pitch are visited 8 times within every 1/15th sec period, yielding 40,328 small detector images per scan frame. The available time for illuminating each focal spot position is divided into multiple short electron beam dwells with cooling in between each dwell in order to increase the tube current limit. At 120 kV, the maximum tube current is 203 mAp (current during each dwell period). In the 71 $\times$ 71 scan mode, the reconstructed field-of-view is 11.4 cm $\times$ 11.4 cm at isocenter. Reconstruction is detailed in Sec. 3.3.

2.2. Image fluence vs. system geometry

The x-ray fluence at a point in a reconstructed image plane scales linearly with the total number of beamlets overlapping that point, and the mean detected fluence per beamlet. As shown in Figure 2, as the scan proceeds along a fixed row of the collimator (defined as the x-direction), the beamlets shift across a fixed point. The total number of beamlets overlapping the point is equal to, on average, the back projected width of the detector

divided by the shift distance between two adjacent back projected detectors. Therefore, for a point at distance z from the source plane, the number of overlapping beamlets in the x -direction is:

$$n_x(z) = \left(W_x \frac{z}{SDD} \right) / \left(\Delta_x \frac{SDD-z}{SDD} \right) = \frac{W_x}{\Delta_x} \left(\frac{SDD}{z} - 1 \right)^{-1} \quad (1)$$

where SDD is source-to-detector distance, W_x is physical detector width along x , and Δ_x is scan pitch at the source plane along the x -direction. Similarly, the number of scan rows overlapping the object along the y -direction is given by

$$n_y(z) = \frac{W_y}{\Delta_y} \left(\frac{SDD}{z} - 1 \right)^{-1} \quad (2)$$

where W_y is the physical detector height and Δ_y is the scan pitch along the y -direction. If the electron beam visits each collimator hole position N_{pass} times in a scan frame, then the total number of beamlets overlapping the point at z is:

$$n_{overlap}(z) = N_{pass} n_x(z) n_y(z) = N_{pass} \frac{W_x W_y}{\Delta_x \Delta_y} \left(\frac{SDD}{z} - 1 \right)^{-2} \quad (3)$$

Equation 3 shows that the number of overlapping beamlets scales linearly with the area of the detector (area = $W_x W_y$). Completing the derivation relating image fluence to system design parameters, we note that the mean fluence at plane z is equal to the sum of the detected fluences from the overlapping beamlets. Using $\varphi_{beamlet}^{SDD}$ to represent the mean fluence rate at the detector from a single beamlet (photons/mm²/s), $t_{capture}$ for the detector capture time for a single beamlet (1.04 μ s), and DQE for detective quantum efficiency, the noise-equivalent image fluence at plane z may be expressed as:

$$\Phi(z) = n_{overlap}(z) t_{capture} DQE \left(\varphi_{beamlet}^{SDD} \right) \left(\frac{SDD}{z} \right)^2 \quad (4)$$

It is assumed that $\varphi_{beamlet}^{SDD}$ is not strongly dependent on the lateral focal spot position and that fluence is relatively uniform over the detector face. The detector capture time per beamlet is constrained by the scan frame period t_{frame} (e.g. 1/15 s), the number of collimator holes (N_x columns by N_y rows), and the number of passes per hole:

$$t_{capture} = \frac{DF \cdot t_{frame}}{N_x N_y N_{pass}} \quad (5)$$

Here the duty factor DF represents the fraction of the frame time that is available for illuminating the collimator holes, rather than moving the electron beam between collimator holes or retracing the electron beam. In the 71 $\times$ 71 15 frame/s mode, $DF = 0.63$. Combining Eqs. (3)–(5) yields the following expression for image fluence at plane z ,

$$\Phi(z) = \frac{W_X W_Y}{N_X \Delta_X N_Y \Delta_Y} DF \cdot t_{frame} DQE(\varphi_{beamlet}^{SDD}) \left(\frac{SDD - z}{SDD} \right)^{-2} \quad (6)$$

The value of the fluence rate $\varphi_{beamlet}^{SDD}$ depends on x-ray tube operating technique (kV, mAs) and patient attenuation. Equation 6 demonstrates that, for a given operating technique and attenuation, the image fluence may be increased by increasing the detector area $W_X W_Y$. By the same reasoning, the primary x-ray exposure at a point in space will also scale with the detector area. (However when considering x-ray exposure in the presence of imperfect beam collimation, the effective x-ray beamlet area at the detector should be used instead of the detector area.) Note that the z-dependence in Eq. (6) represents an inverse square relationship relative to the detector, rather than the source. This dependence is valid for z-values up to the plane where tomographic angle is maximized, at $z = (SDD N_X - X)/(W_X + N_X - X)$.³ We also note that although the quantity N_{pass} does not appear explicitly in Eq. (6), this parameter influences the maximum allowed tube current and therefore the maximum value of $\varphi_{beamlet}^{SDD}$.

Equation (6) suggests there are several ways to improve image fluence. The redesign of the system focused on increasing the fluence by increasing the detector area and accounting for this change in other system components such as the collimator design. With this approach it was possible to revise other aspects of the imaging chain, including reducing the detector element size and changing the reconstructor architecture. As detailed below, an increase in detector area must be balanced against increases in detected scatter, collimator fabrication constraints, and increased data bandwidth requirements. The following section describes the system design.

3. METHODS

3.1. Detector design

To enable imaging with higher fluence, a direct-conversion photon-counting detector array measuring 10.6 cm (horizontal) $\times$ 5.3 cm (vertical) was constructed (Fig 3). As in the previous SBDX detector design, 2-mm thick CdTe was used as the x-ray converter. The detector array has 320 columns $\times$ 160 rows and 0.330 mm wide elements. Full array readout is performed every 1.28 μ s, once per hole illumination. The readout ASIC has a buffer to enable readout of the previous hole illumination while data is being collected for the current hole illumination. Table 1 compares the specifications of the new detector to the previous SBDX prototype.¹ The previous SBDX detector had a 5.5 cm $\times$ 5.5 cm area and used a 48 $\times$ 48 array of 1.14 mm wide elements. The new SBDX detector provides an 86% increase in area and a 71% reduction in detector element width.

As shown in Fig 3A, the detector array was fabricated from 8 $\times$ 4 physically distinct ‘hybrid’ modules. Each hybrid is a low-temperature-solder flip-chip-bonded assembly of a CdTe crystal and a readout ASIC. The modules are arranged in a stepped pattern to accommodate interconnects between the hybrids and a controller board. The carrier structure for the components is water cooled to remove heat generated by the electronics. Each CdTe crystal has one surface of continuous electrode and an 80 $\times$ 80 array of sub-element electrodes on

the opposite surface. A 0.330 mm wide detector element consists of a 2×2 group of binary-counting sub-elements. The ASIC is a custom-designed IC that converts charge collected from the CdTe sub-elements to x-ray counts. Readout circuitry includes a single threshold discriminator and veto logic linking neighboring sub-elements. To avoid double-counting caused by reabsorption of K-fluorescence or charge sharing in neighboring sub-elements, a hit registered by a sub-element leads to deactivation of four neighbors (see Fig 3C). The readout circuitry is designed for the low fluence levels associated with very short sample periods (typically 0 or 1 photon per element per 1 μ s sample period).

Detector size was increased only along the direction of the raster scan lines (x-direction) in order to avoid increasing motion blur. Motion blur may be quantified in terms of the effective pulse width, equal to the time between the first and last beamlet illuminations of a fixed point in the field-of-view. Multi-pass scanning in the source is performed in a blockwise fashion, such that for a 71×71 15 frame/s scan, 8 electron beam passes are performed over a block of 3 scan rows before moving on to the next block. The time to scan one block is

$$t_{block} = N_{pass} N_B (t_{hole} N_x + t_{retrace}) \quad (7)$$

where N_B is the number of rows per block, t_{hole} is the hole-to-hole repetition period (1.28 μ s), and $t_{retrace}$ is the horizontal retrace time. The average number of blocks covering a point is equal to $n_{block}(z) = n_Y(z)/N_B$, giving an effective pulse width of:

$$t_{eff} = n_{block}(z) t_{block} = \frac{W_Y}{\Delta_Y} N_{pass} (t_{hole} N_x + t_{retrace}) \left(\frac{SDD}{z} - 1 \right)^{-1} \quad (8)$$

The calculated effective pulse width is 8.9 ms for a point in the isocenter plane and a 71×71 15 frame/s scan (using $t_{retrace} = 21.8 \mu$ s). From Eq. (8) it can be seen that effective pulse width is independent of detector x-dimension but increases linearly with y-dimension (W_Y).

3.2. Collimator design

A multihole collimator with rectangular apertures was constructed to match the new detector (Fig. 4). The collimator consists of holes that are aimed at the detector and tapered to open up towards the detector. The holes are defined by thin plates of tungsten, brass, and lead with photochemically-etched rectangular holes. As hole depth decreases (moving toward the detector), the hole size increases and the hole pitch decreases. A machined tungsten plate with oversized circular holes provides mechanical support for the thin plates. Strategically placed layers with gaps are used rather than a monolithic design in order to minimize weight and fabrication cost.

Note in the previous collimator design the exit apertures were physically distinct. However with the new wider detector, the horizontal limits of neighboring collimator holes merge together before the exit surface. Therefore the topmost layers of the collimator were designed as slots, and only the deeper layers have physically distinct apertures. The collimator photograph in Fig 4A shows the tungsten slots at the exit surface. The entrance

and exit surfaces of the collimator stack are located 6.2 mm and 50 mm above the target plane, respectively. The multihole collimator is bolted to the face of the scanning x-ray tube. Within the x-ray tube, the thin tungsten target is bonded to a 5-mm thick beryllium plate which forms the exit window of the tube. In the space between the beryllium plate and the entrance of the collimator, there is a ~1.2 mm layer of circulating coolant water and a 0.812-mm thick aluminum x-ray filter.

Simulations of the collimator which accounted for the focal spot size were performed to estimate the spatial distribution of x-rays at the detector plane. Both the new and previous collimator/detector geometries were simulated. We evaluated the ratio of new versus old counts within the detector areas for comparison with relative NEQ measurements. The ratio of new versus old total counts at the detector plane was compared to relative air kerma measurements.

3.3. Real-time image reconstructor

The SBDX images presented for live image display are reconstructed in real time through a two-stage process. First, digital tomosynthesis is used to reconstruct a stack of image planes parallel to the target plane which spans the cardiac volume. The plane-to-plane spacing is selected so that out-of-plane tomographic blurring for a point midway between planes is minimal in comparison to other blurring effects.¹ Second, as described below, a pixel-by-pixel plane selection algorithm is applied to generate a multiplane composite image.

With the larger detector area and smaller detector elements, the data rate from the new detector increased to $(320 \times 160 \text{ elements}) \times (1 \text{ bit per element per hole illumination}) / (1.28 \mu\text{s per hole illumination}) = 40 \text{ Gbps}$. The approximate doubling of detector width also doubled tomographic angle (6 degrees along the x-direction; see Fig 2B), necessitating a reduction in plane-to-plane spacing and an increase in the total number of planes needed to span the imaging volume. To maintain approximately the same depth coverage as the previous system, the number of planes was increased from 16 to 32 and the plane spacing was reduced from 12 mm to 5 mm.

To accommodate the significantly increased data volume and provide a flexible architecture, the reconstructor was redesigned from commercially available GPU-based and FPGA-based components. A flowchart of the imaging chain is shown in Figure 5. Detector data is streamed over two 8-channel fiberoptic cables to the image reconstructor, each carrying 20 Gbps. At the front end of the reconstructor, two FPGA boards (NetFPGA 10G, HiTech Global, LLC) accumulate the detector images that are acquired at a fixed focal spot position within a frame period (e.g. sum 8 detector images per hole, in a 71×71 15 frame/s scan). To reduce downstream bandwidth requirements, 2×2 detector element binning is performed on the FPGA boards. The resulting 160×80 detector images effectively have 0.660 mm wide elements (0.198 mm at isocenter).

The detector images are simultaneously copied to a set of 8 GPUs (Tesla K20, Nvidia Corp.) and two 30 TB real-time disks (Conduant Corp.) through a network of PCI express 2.0 switches (PEX 8680, PLX Technology, Inc.) Together the GPUs perform unfiltered back projection ('shift-and-add' tomosynthesis) at 32 programmed plane positions. The plane

stack reconstructed in a frame period is then copied to two additional Tesla K20 GPUs residing in a Linux host PC. The GPUs in the host PC generate a 2D multiplane composite for live display. Final image processing and display tasks are handled by the host CPU and an Nvidia GTX680 graphics card.

Reconstructor programming was performed in CUDA and C. The reconstruction algorithm has been detailed in Ref. 1 and is briefly summarized here. To generate each tomosynthesis image, the detector images are scaled and summed according to where they back project onto the chosen plane of reconstruction (see Fig. 1). The pixel grid is defined by dividing the shift distance between neighboring back projected images (see Fig. 2A) into an integer number of pixels (10). Typically the 32-plane stack is centered on gantry isocenter and a 5 mm plane-to-plane spacing is used. After plane stack reconstruction, a multiplane composite image is formed to provide a display analogous to a conventional 2D angiogram. First, an adjustable edge-enhancing filter is applied to the plane stack to generate ‘score’ values corresponding to sharp, high contrast features. Then, for each pixel position in the output display, planes with high score values are selected. The displayed pixel value in the composite is a weighted average of the pixel values from the selected tomosynthesis planes.

3.4. Performance evaluation

The image fluence achieved with the new detector and collimator was evaluated with measurements of zero-frequency noise equivalent quanta (NEQ). Acrylic phantoms from 21.0 cm to 35.0 cm thick (20.3 cm $\times$ 20.3 cm lateral dimensions) were centered at gantry isocenter and imaged at 80, 100, and 120 kV using the 71 $\times$ 71, 15 frame/s scan mode. At each kV, imaging was performed at 12.0 kWp (approximately one half full power). Results with the new SBDX system were compared to data from the previous SBDX prototype, detailed in Refs 1 & 2. In cases where the previous NEQ was measured at a different power level, the NEQ was linearly scaled with respect to tube current to obtain a comparable NEQ at 12 kWp. Air kerma rate with the new collimator was measured at the interventional reference point, 15 cm below gantry isocenter. Measurements were performed free-in-air using a calibrated Solidose Model 300 exposure meter.

The NEQ was calculated from the acrylic phantom images using a method employed with the previous SBDX prototype.² First, tomosynthesis images were reconstructed using nearest-neighbor interpolation. In this reconstruction method, the full amount of each detector element sample is assigned to just one reconstructed pixel, resulting in spatially uncorrelated pixel values. A frame-averaged dark-field image acquired with x-rays off was subtracted from the individual flat-field images to remove any deterministic offsets. Next, the mean x-ray signal was measured within a centrally located ROI. The standard deviation of the pixel values in the ROI was calculated from adjacent-frame subtraction images. The standard deviation was divided by $\sqrt{2}$ to account for the subtraction, and then NEQ was calculated as the square of (mean pixel signal/standard deviation). The result, in units of (photons/pixel), was divided by the isocenter pixel area to obtain NEQ in (photons/mm²) at the isocenter plane.

Detected x-ray scatter was evaluated using a standard beam stop method.⁷ An array of 3 mm thick, 2.2 mm diameter lead beam stops embedded in a thin layer of acrylic was placed at

the entrance surface of the acrylic phantom. A frame-averaged image at the plane of the beam stop array was divided by a second frame-averaged image acquired with a thin acrylic 'blank' instead of the beam stops. A dark-field image was subtracted from the numerator and denominator prior to forming the ratio. The scatter fraction value for a beam stop was then calculated as the ratio of the mean value inside the beam stop penumbra divided by the mean value in an annular region around the beam stop. Note the signal behind a beam stop can also arise from off-focus primary radiation, due to x-rays emerging from collimator holes other than the one targeted by the electron beam.² Since this effect depends on x-ray tube details rather than the detector/beam area, a correction was applied to the scatter fraction. Off-focus radiation was evaluated by imaging the beam stops with the acrylic phantom removed and a copper filter placed at the collimator exit.

To demonstrate image reconstruction with clinically realistic anatomy, an image was selected from an ongoing IRB-approved human subject study at the University of Wisconsin – Madison. Similar to a study performed with a previous SBDX prototype,⁸ the SBDX system was temporarily installed beside a conventional clinical fluoroscopic/angiographic system in the University of Wisconsin Hospital and Clinics cardiac cath lab. Subjects requiring diagnostic coronary angiography were screened and consented. The subjects had their standard of care coronary angiogram performed on the conventional x-ray system, and then, using the same catheter placement and with approximately the same contrast dose/flow rate, two SBDX angiograms were acquired.

4. RESULTS

Figure 6A shows zero-frequency NEQ (photons/mm²) at the isocenter plane for the new versus previous SBDX systems, at 120 kV, 12 kWp (71×71 15 frame/s scan mode). Figure 6B shows the comparison at 100 kV, 12 kWp. The kWp refers to the product of kV and the tube current during an electron beam dwell. Full-power tube currents are 203 mAp, 210 mAp, and 209 mAp, respectively, at 120, 100, and 80 kV. At 100–120 kV the NEQ with the new collimator and detector was 63% to 71% higher compared to the previous system, depending on phantom thickness. A similar result was obtained at 80 kV (+60%). For 120 kV 12 kWp operation with a 28.0 cm acrylic phantom, the NEQ of the new system measured 2017.6 photons/mm² at isocenter.

The NEQ comparison depends on relative detector area, as shown in Eq. (6), as well as details of collimation and inherent filtration. Our simulations of the new and previous collimator/detector combinations predicted a 77% (+/-1%) increase in photons falling within the detector area, for 120 kV imaging with 0.6 mm – 0.8 mm focal spot widths. Figure 7A shows an example simulated counts distribution at the detector. A correction must also be applied to account for the fact that the new collimator uses a 0.813 mm Al filter, whereas the previous collimator used 0.406 mmAl. For 80–120 kVp imaging through acrylic phantoms, the calculated reduction in photons at the detector due to the added Al filtration was -3%. Combining the effects of detector area, collimation, and filtration, the expected increase in photons at the detector was $1.77 \times 0.97 = 1.72$, or +72%. The measured change in NEQ was in reasonable agreement with this prediction. The remaining discrepancy may be due to deviations from the collimator design and/or differences in DQE.

To compare x-ray output between the new and previous SBDX systems, the air kerma rate measured at 15 cm below isocenter was normalized by the tube current (mAp). X-ray output on the new system measured 0.775 mGy/min/mAp at 120 kV, 0.511 mGy/min/mAp at 100 kV, and 0.297 mGy/min/mAp at 80 kV. The half value layer of the entrance x-ray beam measured 4.5 mmAl, 3.9 mmAl, and 3.7 mmAl, respectively, at 120, 100, and 80 kV. Compared to the previous SBDX system, the air kerma rate was increased by an average of 53% (57% to 48% depending on kV). The increase in air kerma rate was smaller than that observed in the image NEQ. This may be attributed in part to differences in Al filtration between the two systems, which have a greater impact on the entrance x-ray spectra than the phantom exit spectra due to beam hardening. For identical x-ray energy spectra, simulation of the collimator on each system predicted a 75% increase in air kerma. The calculated effect of the additional Al filtration was a –15% reduction in entrance kerma. Combining the effects yields a predicted kerma ratio of $1.75 \times 0.85 = 1.49$, or +49%.

As anticipated, the improvement to NEQ was accompanied by a modest increase in detected x-ray scatter. X-ray scatter fraction measured 3.9%, 5.4%, and 7.8%, for 23.3 cm, 28.0 cm, and 32.6 cm acrylic phantoms, respectively. The values for 23.3 cm and 28.0 cm phantoms were averaged from 80 to 120 kV; the value at 32.6 cm corresponds to 120 kV. For the same phantom thicknesses and kVs, the previous x-ray scatter fractions measured 2.3%, 3.1%, and 4.2%. Off-focus radiation relative to primary plus scatter measured 6.4% with the new collimator, versus 3.9% with the previous collimator.

Continuous reconstruction of 32 tomosynthetic planes per scan frame plus 1 multiplane composite image per scan frame was successfully achieved when using the $71 \times 71 \times 15$ frame/s scan mode. The initial reconstruction implementation was limited to this scan mode. Figure 8 shows a multiplane composite image recorded by the real-time reconstructor during contrast injection into the left coronary artery. The C-arm was positioned in a posterior-anterior view, and the heart was approximately isocentered. The final display stage of the reconstructor includes an intensity lookup table to reduce image dynamic range. In this example, a logarithmic lookup table was used. Currently either the multiplane composite (Fig. 8A) or a selected single plane image (Fig. 8B) may be viewed live. The single plane display offers the ability to focus on details of a single vessel. Figure 8C demonstrates how two vessels which are crossing in the composite image may be distinguished by viewing different single plane images.

5. DISCUSSION

Scanning-beam digital x-ray is an inverse geometry x-ray fluoroscopic technology under development for cardiac procedures. This work describes a detector, collimator, and reconstructor combination which increases the SBDX image fluence for a given x-ray tube technique. The improvement was achieved by replacing the previous 5.5 cm by 5.5 cm photon counting detector array with a new 10.6 cm wide by 5.3 cm high detector, and increasing the size of the collimator holes to match the new detector. This new geometry increases the number of x-ray beamlets overlapping each point in the field-of-view. The detector size was increased only in the direction of the scan rows (detector width) to avoid increasing motion blur.

Evaluation of the new SBDX system found a 63–71% increase in noise equivalent quanta in reconstructed images and a 47–58% increase in air kerma rate, compared to the previous SBDX prototype. These results were in general agreement with expectations based on simulations of the new detector and collimator. Since detected scatter on the SBDX system is approximately proportional to the volume irradiated at an instant in time, the larger beamlet area resulted in a modest increase in x-ray scatter acceptance. X-ray scatter fraction measured 3.9%–7.8% on the new system versus 2.3%–4.2% on the previous prototype. If desired, other tradeoffs between NEQ improvement and scatter acceptance could be achieved by using differently-sized collimator holes with the new detector.

A new real-time image reconstructor was developed to accommodate the larger, higher-resolution detector. The doubling of tomographic angle in the new SBDX geometry also necessitated a reduction in plane spacing and doubling in the number of simultaneously-reconstructed tomosynthesis planes. The system reconstructs a stack of 32 tomosynthesis planes plus a multiplane composite image for each scan frame, during continuous 15 scan frame/s imaging. In this work, we show an example coronary angiographic image to demonstrate the feasibility of the reconstruction approach. Clinical investigation of angiographic image quality and SBDX radiation dose in comparison to a conventional fluoroscopic/angiographic x-ray system is underway.

The GPU-based reconstruction architecture was adopted to enable programming of new reconstruction techniques as they are developed. This initial implementation was designed for a scan mode using 71×71 focal spots and 15 scan frame/s, even though the x-ray tube is capable of up to 100×100 hole, 30 frame/s imaging. Programming optimizations to enable larger fields-of-view and higher frame rates are underway. Other potential improvements include an increase in the number of planes reconstructed, which would extend the usable imaging volume along the source-detector axis, and the implementation of conventional image post-processing techniques. Finally, we note the reconstructor was designed to support the extraction of 3D information from the tomosynthesis plane stacks. Work to implement frame-by-frame 3D cardiac catheter tracking⁴ in the real-time reconstructor is underway.

6. CONCLUSIONS

The construction of a larger detector and matching collimator for SBDX yielded an increase the noise equivalent quanta and x-ray output for a given x-ray tube operating technique. The modest increase in detected x-ray scatter was in line with expectations. Real-time reconstruction hardware successfully generated 32 tomosynthesis planes plus a multiplane composite image at 15 frame/sec during coronary angiography. These system modifications are an important step towards achieving cine-quality coronary angiographic imaging with SBDX.

Acknowledgments

Financial support provided by NIH Grant 2 R01 HL084022. Technical support provided by Triple Ring Technologies, Inc. and NovaRay Medical Inc.

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Figure 1. SBDX prototype at the UW-Madison (A), and the operating principles of inverse geometry fluoroscopy (B). An electron beam is scanned over an array of positions on a large area target. A multihole collimator between the target and patient defines a series of narrow overlapping x-ray beamlets. The detector images captured in each scan frame are reconstructed at multiple planes using digital tomosynthesis, and a multiplane composite is formed for live display.

Figure 2.
 (A) shows the beamlet shift at plane z when the electron beam advances along a scan row.
 Tomographic angle (θ) is the maximum angular range of rays passing through a point (B).

Figure 3. Photon-counting detector array measuring 10.6 cm by 5.3 cm. The array has 320×160 elements, fabricated from 8×4 distinct ‘hybrid’ modules consisting of a 2 mm CdTe tile bonded to a readout ASIC. The 0.330 mm wide detector elements (orange) each contain 4 sub-elements (dashed lines). Veto logic links each sub-element (green) to four of its neighboring sub-elements (red).

Figure 4. Multihole collimator (A), showing close up of slots on the exit surface (inset). The collimator layers are shown for cross sections perpendicular (B) and parallel (C) to the detector long axis. (Red: tungsten, Green: brass, Dark blue: lead, Gray: aluminum, Light blue: water, Magenta: beryllium.) A single x-ray beamlet is shown in yellow.

Figure 5. Data flow in the reconstructor. Detector images acquired in a scan frame are distributed to GPUs, where a stack of tomosynthesis planes is reconstructed. A multiplane composite is generated from the plane stack for live display. Green blocks are PCIe switches. Right shows the plane stack reconstruction hardware (8 Nvidia K20 GPUs).

Figure 6. Noise equivalent quanta at the isocenter plane for the new and previous SBDX systems, at 120 kV, 12 kWp (A) and at 100 kV, 12 kWp (B). The 71×71 15 frame/s scan mode was used.

Figure 7. Simulation of the x-ray counts distribution within the detector area (A) for the new detector and collimator. A time-integrated detector image is shown in (B). The intensity variations across each tile arise from a gradient in the low energy discriminator threshold across the readout ASIC.

Figure 8. Multiplane composite image recorded during a 15 frame/s coronary angiogram (A) and one of the 32 tomosynthesis planes from the same frame period (B). Vessels that cross in the composite can be distinguished in individual planes (C), as shown for the three images corresponding to the white ROI.

Table 1

Detector specifications

	New	Previous¹
Dimensions (width $\times$ height)	10.6 cm $\times$ 5.3 cm	5.5 cm $\times$ 5.5 cm
Native array size	320 $\times$ 160	48 $\times$ 48
Native element pitch	0.33 mm $\times$ 0.33 mm	1.14 mm $\times$ 1.14 mm
Binned array size/pitch	160 $\times$ 80/0.66 mm	
Capture time	1.04 μ s	1.04 μ s
Native array readout rate	781,250 Hz	781,250 Hz
X-ray converter material	2 mm CdTe	2 mm CdTe