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Inverse geometry x-ray imaging: application in interventional procedures

Michael A. Speidel, PhD

Department of Medical Physics, University of Wisconsin-Madison, Madison, Wisconsin 53705

X-ray fluoroscopy offers features well-suited to image-guided interventional procedures, including continuous real-time imaging, high spatial and temporal resolution, a simple operator interface, and compatibility with all interventional devices. However x-ray dose is an ongoing concern. Additionally, x-ray projection images lack depth information. This is limiting for procedures that require accurate knowledge of catheter position relative to 3D anatomy.

Inverse geometry x-ray fluoroscopy based on Scanning-Beam Digital X-ray (SBDX) technology¹ has potential to reduce x-ray dose in interventional procedures² as well as provide depth information to support image-based 3D catheter tracking,³ tracking of moving anatomic fiducials,⁴ and calibration-free vessel dimension analysis.⁵ This article discusses the potential of inverse geometry x-ray fluoroscopy, SBDX technology, and experimental results obtained with a prototype SBDX system.

INVERSE GEOMETRY X-RAY IMAGING

Inverse geometry x-ray imaging is a method of scanned data acquisition using multiple focal spots distributed over a large area, each of which is collimated to produce a narrow x-ray beam directed at a small detector array (Figure 1). The focal spots are sequentially illuminated and detector images are simultaneously captured, resulting in a series of narrow, overlapping x-ray projections of the imaging volume. The term “inverse geometry” is derived from the fact that the multiple x-ray beams form an x-ray field that is largest at the patient entrance and smallest at the detector. This method of data acquisition has been investigated for fluoroscopy,¹ radiography,⁶ and computed tomography.⁷

Inverse geometry differs from conventional wide-beam projection geometry in three distinct ways:

First, there is a large reduction in detected scatter since only a small volume is irradiated at each instant in time, and the gap between the patient and detector is large. By reducing image-degrading scatter it is possible to achieve a desired object signal-to-noise ratio with fewer primary x-rays and thus lower patient dose. Scatter reduction via beam scanning has been demonstrated in radiography using a moving slit or aperture assembly to define the beam.^{8–10} Non-mechanical inverse geometry scanning extends this well-established principle to rapid, continuous imaging.

ADDRESS FOR CORRESPONDENCE: Michael A. Speidel, PhD, 1005 Wisconsin Institutes for Medical Research, 1111 Highland Ave, Madison, WI 53705, speidel@wisc.edu.

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Second, inverse geometry scanning spreads the primary x-rays over a larger area at the patient entrance. This lowers the average skin dose for a given x-ray output, compared to a stationary conventional system.

Third, imaging in a fixed gantry position is planar tomographic in nature. As shown in Fig. 1B, the rays that pass through each point in the patient arise from a range of source positions. Using digital tomosynthesis, it is possible to reconstruct any focal plane within the patient. The resulting images contain depth information that can be exploited for 3D object localization.

SBDX TECHNOLOGY

Scanning-Beam Digital X-ray is an inverse geometry fluoroscopy system for cardiac procedures (NovaRay, Inc., Newark, CA). The technology aims to reduce x-ray dose while maintaining conventional fluoroscopic and angiographic image quality by (i) minimizing image-degrading x-ray scatter (ii) maintaining high primary detection efficiency at high kVp using a thick CdTe x-ray detector, and (iii) further reducing skin dose for a given integral dose by increasing entrance field area.^{1,2}

Scanning X-ray Source

Inverse geometry x-ray fluoroscopy requires continuous scanning of the patient at up to 30 scans per second. This is achieved by magnetically deflecting an electron beam over a large-area transmission-style tungsten target (Figure 2). The tungsten layer is bonded to a beryllium plate which is cooled with a circulating layer of water. A multihole collimator positioned beyond the target defines each x-ray beam. The electron beam dwells at each hole for 1.04 μ s and moves between adjacent holes in 0.24 μ s.

The SBDX prototype has 100×100 focal spot positions on a 2.3 mm pitch. The source is mounted to a C-arm with a $5.5 \text{ cm} \times 5.5 \text{ cm}$ detector array positioned 150 cm from the source plane, and mechanical isocenter located 45 cm from the source plane. For this system geometry the detected x-ray scatter fraction is 1.8% – 4.6% (18.6 cm – 35.0 cm acrylic phantoms).²

Field-of-view is determined by how many focal spot positions are used in the scan. In the frequently used 71×71 hole, 15 frame/s scanning mode the field-of-view is $11.4 \text{ cm} \times 11.4 \text{ cm}$ at isocenter. The effective pulse width, defined as the time between first and last x-ray beam illuminations of a fixed isocenter point, is 9.4 ms.

One of the challenges of SBDX design is to generate sufficient primary x-ray output to achieve the desired image quality, while staying within the thermal limits of the x-ray tube. The output limitation arises primarily from the use of tight beam collimation, which reduces image scatter but also lowers the number of usable x-rays per electron delivered to the anode. Additionally, differences between SBDX and conventional system geometry require the beam current to be concentrated in a smaller physical focal spot area.

These limitations are offset in part by the fact that the scanning source operates with a higher duty factor for the same effective pulse width compared to a conventional source (63% beam-on time per frame vs. 14%, for 15 frame/s imaging). Furthermore, SBDX employs a multiple-pass scanning technique that reduces peak target temperature and enables higher beam current densities.¹

Source power is expressed as the product of kVp and beam current during a beam dwell (mA). At 120 kVp, the SBDX prototype x-ray source has been continuously operated at up to 24.3 kWp (203 mA). With this beam current and a 0.6 to 0.68 mm diameter circular

focal spot,¹ the beam current density is estimated as 718 to 559 mA/mm². In comparison, the beam current density may be 143 mA/mm² in a conventional angiographic x-ray tube operated in cine mode (123 kVp, 72 kWp, 0.8 mm focal spot, 9° anode angle).

X-ray Detector

Inverse geometry fluoroscopy requires a detector that is both efficient and considerably faster than conventional detector technologies. SBDX uses a 2-mm-thick CdTe photon counting detector that captures images at a rate of 781,250 Hz (one image per focal spot illumination). The capture period is 1.04 μ s and the detector array is read out in the 0.24 μ s period as the electron beam moves between holes.

The prototype detector has 48×48 elements on a 1.14 mm pitch bump-bonded to an integrated circuit readout chip. Due to the very brief illumination period, the number of x-rays incident on the detector surface in a single capture is low (e.g. averaging 0.02 to 2 photons/mm² depending on patient attenuation). Photon counting is performed by dividing each detector element into a number of sub-elements, each of which registers a binary response (0 or 1). Sub-element outputs are then summed to produce the detector element value. The estimated DQE limit for this detector is ~90%, based on modeling of a 2-mm thick CdTe sub-element.²

Image Reconstruction

As scanning proceeds, detector images are transmitted to real-time reconstruction hardware which generates the live fluoroscopic image. First, multiple tomosynthetic images are reconstructed at planes parallel to the source plane (see Fig. 1A). SBDX uses a shift-and-add tomosynthesis algorithm in which the detector images are backprojected and summed at the reconstruction plane. In the SBDX prototype, up to 16 programmable plane positions are reconstructed simultaneously for each frame period (e.g. 16 planes $\times$ 30 frame/s). Each image has the property that in-plane objects appear sharp and out-of-plane objects are increasingly blurred the farther they lie from the reconstructed plane.

For interventional procedures it is desirable for the live image to display all devices and contrast-enhanced vessel anatomy in focus simultaneously. This is achieved by generating a composite image in real-time from the stack of tomosynthetic images. The composite image displays, on a pixel-by-pixel basis, the pixel value from the plane of highest local sharpness and contrast. Sharpness at each pixel position is determined from a gradient-filtered version of the plane stack. The result is analogous to a conventional 2D projection. Figure 3 compares a single tomosynthetic plane and a composite image reconstructed from 16 planes with 12 mm plane-to-plane spacing.

SBDX PROTOTYPE SNR AND EXPOSURE RATE

The iodine signal-to-noise ratio (SNR) and entrance exposure rate of the SBDX prototype relative to a conventional II/CCD-based cardiac system have been reported in Ref ². When the systems were compared at equal kVp and equal SNR versus phantom thickness (23.3 cm to 35.0 cm), SBDX required 69% to 84% lower entrance exposure, with the greatest exposure reduction occurring with the thickest phantoms. These results were consistent with measurements of scatter fraction, detection efficiency, and entrance area versus phantom thickness.

Although the SBDX prototype demonstrated 3 to 7 times higher entrance dose efficiency (SNR²/Entrance Dose), the image SNR of the prototype did not match that of the conventional system at all phantom thicknesses. At maximum x-ray output the SBDX SNR was 107% to 69% of II/CCD cine-mode SNR (18.6 cm to 35.0 cm acrylic). To increase the

maximum achievable SNR, the system is being redesigned to double the width of the x-ray beam and detector (along the direction of the scan rows). This will double the number of beams overlapping each point in the patient, and is projected to double the image x-ray fluence for a given source power.

3D CATHETER TRACKING

The rapid tomosynthesis capability of SBDX offers the unique opportunity to image cardiac anatomy and catheter devices with depth resolution in real time. Recent work has shown that SBDX depth resolution can be exploited to perform precise 3D localization of metal catheter electrodes.³ 3D catheter tracking is useful in procedures such as catheter ablation of cardiac arrhythmia, where the interventionist is navigating the catheter tip to specific surface sites inside a cardiac chamber.

SBDX 3D catheter tracking makes use of the tomosynthetic reconstructions and gradient filtering techniques developed for composite image reconstruction. The tracking algorithm segments the gradient-filtered image stack into distinct object regions. Then, for each object region a 3D coordinate is calculated. Tracking performance has been investigated for cardiac catheters in phantoms.³ Results show single-frame 3D localization accuracy and precision of 1 mm or better can be obtained.

The proposed tracking method provides 3D positions at the rate of fluoroscopic imaging, uses a fixed gantry angle, works with standard catheters, and requires no geometric calibration. Future work is focused on a real-time implementation (a clinical requirement) and methods for registering tracked coordinates with preacquired CT and MRI volume images.

CONCLUSIONS

Inverse geometry x-ray imaging is a rapid narrow-beam scanning technique which carries the benefits of scatter reduction, large entrance area, and imaging with depth resolution. The SBDX system uses inverse geometry for the purpose of dose reduction in fluoroscopy and cardiac angiography. Design challenges include achieving sufficient x-ray source output, fast readout rates in the detector, and the need for real-time image reconstruction. Results with the SBDX prototype system indicate this technology may enable large dose reduction in interventional procedures, and also support novel interventional tools based on real-time tomosynthesis.

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Figure 1. Inverse geometry x-ray imaging. **A:** Focal spots on a source plane are sequentially illuminated to generate a narrow scanning x-ray beam directed at a small detector. **B:** Each point in the patient is imaged from a range of focal spot positions, allowing object depth to be determined.

Figure 2.
SBDX scanning x-ray source with thin transmission target and narrow-beam multihole collimator.

Figure 3.
A: Single plane tomosynthesis reconstruction of a cardiac chest phantom. **B:** Multiplane composite reconstruction of the same phantom (scanning was performed in 71×71 hole, 15 frame/sec mode).