The feasibility of an inverse geometry CT system with stationary source arrays

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Purpose: Inverse geometry computed tomography (IGCT) has been proposed as a new system architecture that combines a small detector with a large, distributed source. This geometry can suppress cone-beam artifacts, reduce scatter, and increase dose efficiency. However, the temporal resolution of IGCT is still limited by the gantry rotation time. Large reductions in rotation time are in turn difficult due to the large source array and associated power electronics. We examine the feasibility of using stationary source arrays for IGCT in order to achieve better temporal resolution. We anticipate that multiple source arrays are necessary, with each source array physically separated from adjacent ones.

Methods: Key feasibility issues include spatial resolution, artifacts, flux, noise, collimation, and system timing clashes. The separation between the different source arrays leads to missing views, complicating reconstruction. For the special case of three source arrays, a two-stage reconstruction algorithm is used to estimate the missing views. Collimation is achieved using a rotating collimator with a small number of holes. A set of equally spaced source spots are designated on the source arrays, and a source spot is energized when a collimator hole is aligned with it. System timing clashes occur when multiple source spots are scheduled to be energized simultaneously. We examine flux considerations to evaluate whether sufficient flux is available for clinical applications.

Results: The two-stage reconstruction algorithm suppresses cone-beam artifacts while maintaining resolution and noise characteristics comparable to standard third generation systems. The residual artifacts are much smaller in magnitude than the cone-beam artifacts eliminated. A mathematical condition is given relating collimator hole locations and the number of virtual source spots for which system timing clashes are avoided. With optimization, sufficient flux may be achieved for many clinical applications.

Conclusions: IGCT with stationary source arrays could be an imaging platform potentially capable of imaging a complete 16-cm thick volume within a tenth of a second. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4789918>]

Key words: inverse-geometry CT, stationary sources, volumetric reconstruction

I. INTRODUCTION

In clinical practice, computed tomography (CT) scanners have often battled against time. Patient motion is inevitable, and the most reliable means for reducing motion artifacts has always been to scan faster. The first three generations of CT saw three different visions of data collection,¹ with each architecture enabling new applications by virtue of being significantly faster than its predecessor. Early attempts for extremely high-speed imaging include the dynamic spa-

tial reconstructor, an impressive prototype CT platform with 14 x-ray sources and cameras and also electron beam CT (EBCT).^{2,3} Conventional systems improved their temporal resolution by reducing the gantry rotation times, but substantial further decreases in the rotation time are challenging because of the centripetal forces exerted on all the gantry components and especially the fragile x-ray source. CT systems have achieved a tremendous increase in the number of slices imaged per unit time using an orthogonal approach, through multislice CT (MSCT),^{4,5} but the temporal resolution has

decreased more slowly. Applications that require freezing cardiac related motion demand extremely fast temporal resolution and are ultimately limited by the gantry rotation time.

Dual source CT achieves an improvement in temporal resolution of a factor of 2 by mounting two source-detector assemblies on the same gantry.⁶ It is highly unlikely that the dual source CT concept can be expanded to fit even more source-detector modules for further scan time reductions. Indeed, in the current commercial implementation, one of the detector arrays has a limited field of view due to mechanical constraints.

An alternative to rotating the gantry faster is to avoid rotating the gantry at all. EBCT was developed upon this premise, and it featured a very large, stationary source array in conjunction with a large, stationary detector that could not be coplanar. EBCT was successful at achieving very high temporal resolution and for some time found use in cardiac applications.³ However, EBCT also suffered from a range of problems including modest spatial and contrast resolution, artifacts from high scatter and the use of noncoplanar rays, and it could not easily be extended to cover large volumes. As a result, as conventional systems became faster, EBCT was replaced by systems with mechanical rotation.

Scanners that image hundreds of slices in parallel using a cone-beam geometry are a reality,⁷ but they are starting to hit limits. Some applications, such as whole-organ perfusion, require the capability to image large volumes quickly with high image quality and quantitative accuracy. Avoiding cone-beam artifacts with a single axial scan, however, is fundamentally impossible with cone-beam systems because of data insufficiency.

Recently, inverse geometry CT (IGCT) was proposed as a CT architecture with several potential advantages. The fundamental idea of IGCT is to use a small detector and a distributed source array.^{8,9} Use of multiple sources in the axial direction solves the data insufficiency problem and thereby eliminates cone-beam artifacts. The small size of the IGCT detector also significantly reduces scatter acceptance, and may enable technologies such as photon counting detectors. With a traditional system, the modulation of the source current (*tube current modulation*) is able to optimize the incident flux as a function of view direction to increase the dose efficiency.¹⁰ With IGCT, the use of multiple sources illuminating a small detector allows for further control of the photon flux. Modulating the output of each source gives us the ability to dynamically control the distribution of flux within a fan (*virtual bowtie*). The system now has the flexibility to send more photons through rays that have high attenuation or otherwise are clinically important while sparing tissues that are less clinically relevant or may be more radiation sensitive.¹¹ Figure 1 compares conventional CT with IGCT.

Three variants of IGCT have been described. The *original* IGCT used a large, dense, distributed source array and a small x-ray detector.⁹ The nature of the inverted geometry suggests that the size of the source array in IGCT should be comparable to the size of the detector in conventional systems. Such a large source array may be difficult to build in practice. To reduce the size of the source arrays, the *multidetector* IGCT was

proposed.¹² Separately, the *multispot* IGCT was proposed and then constructed for experimental use. The multisource IGCT uses a smaller number of discrete x-ray emitters and can be viewed as intermediate between the conventional CT system and the standard IGCT system.^{13,14}

While IGCT has significant potential advantages over conventional scanners, it also has several limitations. The distributed source array in IGCT is much more complex than the traditional x-ray tube and also quite large, even with the use of multiple detectors. The source array and associated high power and high voltage supplies (and possibly vacuum pumps) are extremely difficult to rotate quickly because of their large size and complexity. Achieving temporal resolutions comparable to current third-generation designs will be extremely challenging. Another disadvantage of IGCT is its limited useable flux, owing to the small size of the detector as compared to traditional systems. IGCT is able to make better use of its flux because of virtual bowtie, but acceptable flux and noise levels in clinical applications have yet to be demonstrated.

To overcome at least some of these limitations, we propose a stationary source IGCT (SS-IGCT) system that would comprise a ring of stationary source arrays. Only the small detector array and a collimator array need to rotate, thereby eliminating the major obstacle to faster gantry rotation. In spirit, this architecture combines the concepts from IGCT, MSCT, and EBCT. In our design, the IGCT concept of a small rotating detector overcomes the EBCT drawbacks of scatter and the data insufficiency problems of both EBCT and cone-beam MSCT. The EBCT concept of stationary sources allows for very high temporal resolutions that would be difficult with mechanically rotated sources. Finally, the large number of detector rows allows wide axial coverage in a single rotation. In contrast to EBCT, our system still uses a rotating detector and collimator, but we anticipate that without the fragile x-ray source, the new gantry can complete a revolution within a tenth of a second.

The scanner is designed for fast, volumetric imaging. Potential clinical applications include cardiac imaging, dynamic whole-organ imaging such as brain or liver perfusion studies, and pediatric imaging with reduced or no sedation. Cardiac imaging is one of the most demanding applications in CT with challenging requirements in temporal resolution, spatial resolution, and volumetric coverage. A variety of unconventional architectures are currently being investigated to meet the stringent requirements for cardiac scanning.¹⁵ Our target performance requirements include 100 ms temporal resolution, 16 cm of coverage in the axial direction, and spatial resolution comparable to existing scanners. Conventional volumetric scanners capable of imaging the heart in a single rotation use large cone angles, leading to fundamental data insufficiency and cone-beam artifacts. An additional requirement on the system is to reduce the level of cone-beam artifacts to a level comparable to that which is currently tolerated in scanners with 4 cm of axial coverage. Another design consideration that is particularly relevant for IGCT scanners is flux. We will discuss the flux requirements for the scanner and suggest that flux sufficient for many clinical applications is plausible.

FIG. 1. Conventional CT, IGCT, and the proposed SS-IGCT. (a) Standard, third-generation CT uses a rotating source and detector to collect a fan beam of data. (b) IGCT inverts the geometry by using a small detector and a series of sources opposite the detector. A fan-like shape of data, analogous to the fan-beam of conventional CT, is collected. Note that these x-ray beams are not illuminated simultaneously, but instead in sequence. The requirements on the rotating source array are very demanding, and the source array itself is very large. (c) The proposed system, SS-IGCT, increases temporal resolution and avoids the complexity of rotating source arrays by using three stationary source arrays. The gaps between the source arrays lead to missing data. Note that the inverted fan beam, beginning at the detector and ending at the source arrays, is missing a portion of the data because of the gap. These missing data prevent the direct use of conventional reconstruction algorithms. (d) The SS-IGCT system shown with additional hardware. In this schematic, the stationary source array is implemented using electron guns. The collimator and detector rotate together. Only a single x-ray beam is shown, corresponding to a single moment in time. The FOV here is the scanner field of view. [Reprinted with permission from S. S. Hsieh and N. J. Pelc, "A volumetric reconstruction algorithm for stationary source inverse-geometry CT," *Proc. SPIE 8313, Medical Imaging 2012: Physics of Medical Imaging*, 83133N (2012)].

Other CT systems using source arrays have been proposed. One concept uses a helix-shaped stationary source array and a large stationary detector with a helical gap. The combination allows for fast, volumetric imaging, but at the cost of some missing data.¹⁶ A line source array has also been proposed for large axial coverage, but requires the assumption of periodic motion in order to achieve high temporal resolution.¹⁷ A "semistationary" design using multiple point sources on a rotating gantry has also been proposed, which uses interior tomography algorithms with the goal of achieving high temporal resolution in cardiac applications.¹⁸ The proposed SS-IGCT system differs from previous work in being able to directly measure a complete dataset with high temporal resolution.

In order to demonstrate the feasibility of our architecture, we will first present the basic system concepts and one system design that we believe can be physically realized with existing technology, and then demonstrate that the image quality of the system is acceptable. Specifically, we will demonstrate that the spatial resolution and photon utilization efficiency of the proposed system are comparable to existing scanners and that the magnitude of the artifacts is smaller. We will not ad-

dress issues of cost in this work. The most fundamental problems of reconstruction and timing will be presented, followed by potential solutions. We will not discuss in detail the actual implementation of the source array, although the most obvious candidates for such a stationary source array would be electron beam systems¹⁹ and carbon nanotube-based source arrays.^{20,21} Because electron beam source arrays are relatively mature, have small source spacing, and have demonstrated long lifetime and high instantaneous power, we will assume the use of steered electron beams as in the scanning beam digital x-ray (SBDX) system.¹⁹

The x-ray sources need to be in high vacuum yet the source housing also needs a relatively x-ray transparent surface facing the patient. Because of mechanical difficulties in constructing a large ring vacuum chamber with these capabilities, we anticipate that we will need to use a series of separated source arrays, with gaps between them. These gaps produce missing data in the sinogram, but we provide an algorithm that is able to eliminate most of the resulting artifacts. With a moving collimator and detector, the firing of the individual x-ray source-spots requires careful timing to ensure the collection of the needed data with no timing clashes, which

would occur if two sources were scheduled to fire in the same time interval. We provide a potential timing scheme for our system that eliminates pulse timing clashes. Finally, we simulate the system to ensure that the noise and resolution characteristics remain intact and that the artifacts produced by the gaps are minimal.

II. SYSTEM DESIGN

The design goal of our system is to provide a dataset equivalent to that of a third generation scanner with multiple rotations, but to do so with stationary source arrays and a single rotation of a small detector. The stationary source arrays allow for a reduction in gantry rotation time while the small detector captures the benefits of IGCT. To simplify the discussion, we will initially focus on the stationary source arrays and tacitly assume a large detector. The small detector will be introduced later in order to complete the system. Figure 1 shows two-dimensional schematics for the SS-IGCT system in comparison to conventional CT and standard IGCT systems.

We assume throughout this work that the scanner produces an image from a single axial rotation, without table translation. Although it would be possible to generalize the design and reconstruction to the helical scanning mode, we focus only on the axial scan, which is most relevant for many clinical applications such as whole-organ perfusion or cardiac scanning.

II.A. Source design

The simplest stationary source array design would be a single, monolithic, cylindrical source array that completely encloses the patient. This source array would be capable of mimicking the action of any traditional source trajectory; instead of the x-ray source rotating, source spots would fire in a circular (or helical) manner that would correspond exactly to the motion of the source. The collimator would rotate directly inside the source cylinder and pass over each source as it is energized, and the detector (assumed for now to have a full width) would rotate directly opposite the collimator. We have removed from the rotating gantry the x-ray source, the high voltage generator, any needed vacuum pumps, and the high power electronics. Note also that since the source components are stationary the slip ring must only transmit power and control signals to the detector electronics and digital data from the detector. We therefore expect that this system could have extremely good temporal resolution. The source spacing would be small enough to ensure a sufficient number of views. This is on the order of millimeters, which is achievable with a scanned electron beam source (SBDX).

Although simple in concept, this design may not be mechanically feasible. The engineering problems associated with building a source array scale poorly with size, and additionally, a single large source array is undesirable from the standpoint of engineering modularity. Therefore, we propose breaking up the single large source array into a series of separated source arrays. The multiple source arrays act identically to the large, monolithic unit, except at the transitions between

the source arrays. At these locations, a gap between source arrays seems inevitable. These physical gaps will produce missing views in the sinogram. We assume in this work that the source gaps are parallel to the central axis of the system.

In single slice (2D) imaging, estimates for these missing rays can be found using symmetry, because every line integral is measured twice in a full 360° scan. We will use the term *equivalent symmetric ray* to refer to the ray that measures exactly the same line integral, which is nominally a perfect substitute. We expect that missing rays are tolerable if the equivalent symmetric ray is present. With discrete source positions and a band-limited object, interpolation can be used to estimate missing rays as long as the number of source positions is sufficiently large. With a series of stationary source arrays, missing rays, which originate in a gap but in which the other end point is not in a gap, have an equivalent symmetric ray. However, the line integrals connecting one gap to another gap are truly missing rays because neither of the two equivalent symmetric rays are measured. These truly missing rays are unrecoverable via symmetry.

The solution that we propose is to use three stationary source arrays. In this case, the truly missing rays define a triangle, and if the object is fully contained within the triangle, then the truly missing rays are known to be zero. Note that if an even number of source arrays is used, truly missing rays will pass through the isocenter. If the number of source arrays is an odd number greater than three, truly missing rays connecting nonadjacent gaps pass fairly close to isocenter. The choice of using three source arrays, therefore, provides the largest field of view with no truly missing rays. Figure 2 shows the triangular field of view.

II.B. Volumetric design

A key advantage of stationary source arrays is in volumetric reconstruction. Third generation scanners must accept a trade-off between cone-beam artifacts and motion effects. Using a single axial scan, traditional scanners can quickly obtain large volumetric coverage, but with correspondingly large cone angles and hence large cone-beam artifacts. Alternatively, a spiral trajectory could be used with a smaller cone angle, avoiding cone-beam artifacts but increasing susceptibility to motion inconsistencies throughout the volume. Stationary source arrays allow us to acquire the equivalent data of several axial scans in parallel with one rotation of the gantry. We will say that the source points that generate each separate axial scan constitute a *source row*. In other words, a source row is a ring of source points with the same value of z (the axial direction), and all of the sources in a single source row together produce the dataset for a single axial scan. Note that the stationary source can also achieve source trajectories that are more complicated than the multiple rings.^{22–24} These trajectories may present advantages over the traditional axial scan, but in this present work we limit ourselves to multiple rings to simplify the reconstruction problem. Figure 2 shows a volumetric rendering of the proposed scanner, with the multi-ring source trajectory highlighted.

FIG. 2. Volumetric depiction of SS-IGCT, showing the detector, source arrays, source trajectory and the x-ray beam. The five source rows, which constitute the source trajectory of the system, are drawn on the source array as circular stripes. Some elements, such as the collimator, have been omitted in this schematic. (a) Two x-ray beams are emitted from two different source rows. (b) The triangular field of view is shown.

To reduce the cone-beam artifacts, it may seem that collimating out rays with a significant tilt angle with respect to axial slices would be desirable. However, this would reduce the flux of the system if the scan time is kept constant. A variant of FDK has been proposed that is designed specifically for multiple parallel axial scans, and makes use of rays with large tilt angles without introducing cone-beam artifacts.²⁵ It is therefore not necessary to collimate out rays with a large tilt angle, and there is no reduction in flux for using multiple source rows in parallel compared to using a single source row.

While the stationary source arrays allow the use of multiple source rows, they also introduce missing rays as described above. As previously mentioned, in 2D imaging, symmetry gives us the ability to replace all missing rays with their equivalent symmetric rays as long as the object is contained within the triangle defined by the three gaps. In three dimensions, this is no longer true. The sampling of the source array in the z direction is sparse if we only have sources that are contained along several source rows. The sampling on the detector, in contrast, is dense. Our proposed system parameters (Table III) include 160 detector rows but only 5 source rows. Had the sampling on the source array been dense, we would use symmetric equivalent rays, but with coarsely spaced source rows the symmetric equivalent ray will usually not be available.

If we could accurately estimate the missing rays, we would have complete data for multiple parallel axial scans, and we could obtain a high-fidelity volumetric reconstruction using existing algorithms. Conversely, if a high-fidelity volumetric reconstruction were available, we could easily estimate the missing rays using forward projection. With neither an estimate of the missing rays nor a good initial reconstruction, we consider an iterative approach that successively obtains more accurate estimates of the missing rays and the corresponding volumetric reconstruction. We will present a method that in our simulation studies requires only a single iteration and which may therefore be computationally more efficient than

methods which require several forward projections and back projections. We begin by using a subset of the measured data to produce a *starting reconstruction* that is noisy but sufficiently accurate. Forward projection is then used to estimate the missing rays, and a *final reconstruction* is then generated from the completed dataset. Additional iterations could be used to further improve the estimate of the missing data. Empirically, we find that with the proposed starting reconstruction additional iterations are not necessary.

II.C. Starting reconstruction

For the starting reconstruction, we ignore rays that have a large tilt angle, as if they had been collimated out. This increases the accuracy of the starting reconstruction. While discarding these rays greatly reduces the effective flux and hence increases the noise of the starting reconstruction, only the missing data, which is a small fraction of the full dataset, will be estimated from the starting reconstruction. With the system parameters proposed below, the missing rays constitute about 5% of the total rays. Hence, for the purpose of our starting reconstruction, we ignore rays with large tilt angle and accept the resulting noise penalty in favor of accuracy.

Consider the reconstruction of a voxel (the “voxel of interest”), which is close to a plane containing a source row (that is, with a small tilt angle). For now, let the voxel lie on the central axis of the system. Because of the gaps, several of the rays that would otherwise measure line integrals passing through the voxel of interest are missing. The number of missing rays is equal to the number of source spots that would have been in the gaps. If the voxel of interest is exactly on a source plane, the tilt angle is zero and we have a perfect substitute with the equivalent symmetric ray. With a nonzero but small tilt angle, a reasonable substitute for the missing ray is the *conjugate ray*. The conjugate ray is symmetric to the missing ray; it emanates from the source spot on the same source row but opposite the missing spot and ends on the same

FIG. 3. (a) The voxel-by-voxel conjugate ray. Typically, the standard conjugate ray would be used as a substitute for the missing ray. In the starting reconstruction, we choose to use a voxel-by-voxel conjugate ray. For Voxel A, which lies on the axis of rotation (the black dotted line), the voxel-by-voxel conjugate ray is equivalent to the standard conjugate ray. For Voxel B, which lies off the axis of rotation, the voxel-by-voxel conjugate ray is coplanar to the missing ray and the standard conjugate ray but also passes through Voxel B. (b) Reconstruction using the standard conjugate ray as a substitute for the missing ray. A longitudinal slice is shown of four water spheres, each 1 cm in diameter and 10 cm from the isocenter. The top sphere is in the plane of the source row, and the bottom sphere is 4 cm from the isocenter. (c) Reconstruction using the voxel-by-voxel conjugate ray as a substitute for the missing ray. The voxel-by-voxel conjugate ray reduces artifacts and the faint, ghost spheres that appeared with the standard conjugate ray. [WL, WW] = [−1000, 80].

detector row as the missing ray. Figure 3 shows a voxel on the central axis, a missing ray, and a conjugate ray. The conjugate ray does not measure the same data as the missing ray except as the tilt angle goes to zero when it becomes the equivalent symmetric ray.

When the voxel of interest is not on the central axis, there are two obvious choices for the conjugate ray. One choice is to continue as if the voxel were still on the axis of the system, i.e., using the same source and detector row as the missing data. This choice is appealing from the perspective of simplicity: we start from a dataset that has missing rays, replace each missing ray using a conjugate ray to complete the dataset, and pass the completed dataset directly to an algorithm such as FDK to obtain the initial reconstruction.²⁶ An alternative choice is to choose another conjugate ray that actually passes through the voxel of interest. This is more accurate but at first glance appears to be much more complicated. On a voxel-by-voxel basis, it seems we must select a new set of conjugate rays. In fact, choosing the conjugate rays on a voxel-by-voxel basis, if done after the filtering step, is not very difficult, the difference being a single lookup operation so that the incremental computational cost is minimal. In our scheme, the

filtering is done before the lookup operation, so that no additional ramp filtering steps are necessary. We will therefore use the second choice; that is, we will use the voxel-by-voxel conjugate ray, ensuring that every conjugate ray we use passes through the voxel of interest. Figure 3 shows a voxel off of the central axis and the voxel-by-voxel conjugate ray used to reconstruct it, and also an example reconstruction using the standard conjugate ray and the voxel-by-voxel substitution.

Mathematically, let $g_f(\alpha, \gamma, \theta)$ be filtered projection data, where α is the tilt angle, γ is the inplane fan angle, and θ is the rotation angle. Depending on the scanner geometry, $g_f(\alpha, \gamma, \theta)$ may also include a density correction factor. Most algorithms of the FBP form (including most notably the FDK algorithm) specify a reconstruction of $f(\vec{x})$ using the following formula:

$$f(\vec{x}) = \int_0^{2\pi} c(\vec{x}, \theta) g_f(\alpha(\vec{x}, \theta), \gamma(\vec{x}, \theta), \theta) d\theta.$$

Here, $c(\vec{x}, \theta)$ includes all other factors in the FBP reconstruction, and $\alpha(\vec{x}, \theta)$ and $\gamma(\vec{x}, \theta)$ are functions which select the correct tilt angle and fan angle in the dataset. The details of these functions do not affect our method of missing data compensation and are omitted to simplify the discussion. The interested reader is referred to Ref. 27 for a more indepth description. Because of the gaps, $g_f(\alpha, \gamma, \theta)$ is not known for certain missing views, and we must therefore choose a substitute. Let $\beta(\theta, \gamma)$ be the view angle of the conjugate ray if θ is a missing view, and equal to θ if θ is measured. We propose to use as a starting reconstruction a modified FDK formula

$$f(\vec{x}) = \int_0^{2\pi} c(\vec{x}, \theta) g_f(\alpha(\vec{x}, \beta(\theta, \gamma)), \gamma(\vec{x}, \beta(\theta, \gamma)), \beta(\theta, \gamma)) d\theta.$$

According to this formula, if the view is not missing, then the FBP algorithm proceeds as usual for that view. If the view is missing, then the view corresponding to the conjugate ray, $\beta(\theta, \gamma)$, is instead used. The scaling factor $c(\vec{x}, \theta)$ remains unchanged.

Our system uses several source rows (each producing a separate dataset), which can in turn be converted to separate reconstructions. The technique described above for the starting reconstruction assumed the cone angle was small. One approach for a thick volume reconstruction is to define axial slabs centered on each source row and which together cover the entire field of view, and to use each source row only to reconstruct its corresponding axial slab. However, such an approach is likely to generate artifacts at the transitions between slabs. Therefore, we apply a linear blending in z to the reconstructions to reduce the artifacts. This completes the starting reconstruction.

From the starting reconstruction, a forward projection is used to estimate the missing data, making our dataset complete. An existing algorithm is then used to produce the final reconstruction using all the data, not just the rays with narrow tilt angle.²⁵

Other volumetric reconstruction approaches are possible. Super-short-scan reconstruction algorithms, for example, can be applied quite directly to reconstruction problems with gaps.^{28,29} These algorithms are less conventional and their artifacts (in particular, their cone-beam artifacts) are not as well known, but they remain an attractive approach. While we believe that the reconstruction algorithm presented here is of sufficient quality and demonstrates the feasibility of the system concept, more work is necessary to explore and characterize alternative and potentially superior reconstruction approaches.

II.D. Detector and collimator design

We now turn our attention to the use of the small detector, standard in IGCT designs and potentially capable of reducing the scatter and increasing the dose efficiency.

It is common to model scatter as a slowly spatially varying intensity that is added to the detected primary radiation. Both in this model and in experiments, the total scatter to primary ratio is approximately proportional to the beam area, and therefore the total detector area.³⁰ With the small detector of IGCT, the scatter to primary ratio is expected to be several times less than that for conventional scanners with the same axial coverage. We believe that this eliminates the need for an antiscatter grid from the system. Without the antiscatter grid, the geometric efficiency of the detector can be improved. Aside from controlling the scatter to primary ratio, the size of the detector also impacts the available flux in the system and the ability of the virtual bowtie to customize the flux profile to the clinical task.

Consider the data that needs to be collected from a single source spot. We would like to use a small detector to produce the same data that would have been acquired with a large detector. Using the large detector, the source spot would only be energized once when the detector is directly opposite it. With a small detector, this source spot must be energized multiple

times as the small detector rotates. The small detector snapshots can be aggregated and interpolated to produce a single view, with essentially the same data as if we had used a large detector. Indeed, the geometry of snapshots can be arranged to achieve oversampling in the radial sinogram direction and avoid aliasing artifacts.^{31,32}

For this aggregation process to be truly seamless, the detector must be an arc of a cylinder centered on the central axis. This differs from both the traditional flat detector and the equiangular detector that has a radius of curvature equal to the source-to-detector distance. A large detector that is an arc about isocenter can be perfectly mimicked, without any interpolation concerns, by several small detector snapshots. If the length of the detector is l and the detector-isocenter distance is DID, then a large detector (without oversampling) can be built up by taking a detector snapshot, rotating the detector by $\frac{l}{\text{DID}}$ rad, and repeating the process as needed.

To be precise, let $\Phi = \{\phi_1, \phi_2, \dots, \phi_M\}$ denote the set of gantry angles for which the single stationary source spot will be energized with $\phi_i = 0$ defined as the angle at which the detector is directly opposite the source spot. Let us make clear that the cardinality of Φ , M , is the number of times this source spot will be energized. If we chose $M = 3$, then one obvious choice is $\Phi = \{-\frac{l}{\text{SDD}}, 0, \frac{l}{\text{SDD}}\}$, where SDD denotes the source-to-detector distance. This choice of Φ builds up an effective detector that is three times larger than the actual small detector, and is shown in Fig. 4. In general, this obvious choice for Φ may not be optimal. The angles Φ must be chosen so that the source spot can stitch together an equivalent large detector without missing data, but they may additionally include other constraints. To prevent aliasing, we may oversample the angles Φ . To achieve greater fluence near the center of the patient, where attenuation is highest, we may choose Φ to be nonuniformly spaced. For reasons such as these, we will work with a general set Φ rather than a specific example.

It is clearly important to design a collimator so that only photons directed at the detector array are transmitted. We will use a collimator that rotates synchronously with the detector.

FIG. 4. Detector stitching, the collimator, and the set of angles Φ . (a) A conventional system is shown with a large detector, collimator, and x-ray source emitting a fan beam of radiation directed toward the detector. (b) The SS-IGCT system is shown with a smaller detector and the collimator. The collimator (drawn thick) lies just within the source ring, and the collimator itself has three holes. To minimize confusion, gaps between source arrays are not shown. We use $\Phi = \{-\frac{l}{\text{SDD}}, 0, \frac{l}{\text{SDD}}\}$. A source spot lies on one of the three stationary source arrays, which illuminates the detector through a hole in the collimator. (c) and (d) The collimator and detector have rotated such that another collimator hole is present in front of the same source spot. The source spot is re-energized. The data collected from the three source firings, illustrated in (b)–(d), are equivalent to the single image with the large detector in (a). The thickness of the collimator has been exaggerated in these figures to show more clearly the directionality of each collimator hole, which allows the collimator to filter out radiation which is not directed toward the detector.

The collimator has many holes, and the locations of these holes are simply given by $-\Phi$. Under this design, when a source spot wants to fire for the i th time, the detector has rotated to ϕ_i rad away from being directly opposite, and the i th collimator hole has likewise rotated into position right in front of the source spot. The same collimator holes are used by the other sources in the same ring, and the timing of firings of the other sources is simply offset by the angular positions of the sources.

To summarize, each source spot, together with the smaller detector, will take M snapshots, with the detector rotated between snapshots. The relative location of the detector at each different snapshot is parameterized by the set of angles Φ . The collimator consists of M holes, with the location of the holes given by $-\Phi$. As the gantry completes one revolution, all M holes will pass over each source spot, and when they coincide the source spot will be energized. Afterwards, the data from all M snapshots can be aggregated to produce a virtual large detector.

The problem now becomes that of scheduling. How do we ensure that there are no timing conflicts? With multiple source spots, it could be the case that two source spots are scheduled to fire at the same time, a request that would be impossible for the system to fulfill.

II.E. Timing

Under the optimal timing scheme, we must have only one source spot energized at any given time, and ideally we would like one source spot energized at all times so that the system does not have any idle time.

For simplicity, we begin by considering a single slice (two-dimensional) system. Let the system contain N views, or equivalently, N source spots that are spaced at equiangular points on the source arrays. Included in these N views are source spots that are missing because they are within gaps but our timing discussion will proceed as if there were no gaps. Each of these N source spots will take M snapshots with the small detector, with the relative location of these snapshots parameterized by Φ . We impose the requirement that Φ is the same from source to source (i.e., all sources in the ring use the same collimator holes).

In one 2π rotation of the gantry, there are a total of NM source firings. Accordingly, we divide the 2π rotation into NM angular slots, with one slot being a rotation of $2\pi/NM$ rad. The slot is a unit of angle but also indirectly of time, because the rotation of the gantry by one slot will correspond to a fixed interval of time. Our goal is to associate each of the NM source firings with a unique gantry position or time slot.

We use $K = \{k_1, k_2, \dots, k_m\}$ to denote the collimator locations Φ , converted from radians into slots, which we round to the nearest integer. (The number of slots is generally large enough so that the impact of this rounding error is minimal.) Because we are working with a circle, let $K = K \bmod NM$ so that $0 \leq k_i \leq NM - 1$. When the gantry has rotated by T slots, the collimator too will have rotated by T slots, and the new location of the collimator holes will be at $(K + T) \bmod NM$.

The source spots are spaced equidistantly on the source arrays, so they can also be described as residing in slots $\{0, M, 2M, 3M, \dots, (N - 1)M\}$. These source spots are fixed and do not move as the gantry rotates. As we stated previously, whenever a collimator hole coincides with a source spot, the source spot energizes. This rule ensures that each source spot acquires a complete dataset, equivalent to that of a larger detector.

We note that the N source spots, located at slots $\{0, M, 2M, 3M, \dots, (N - 1)M\}$, are all zero modulo M . On the other hand, when the gantry is in position T , the M collimator holes are at $(K + T) \bmod M$. In order to avoid timing clashes, we need the M collimator hole starting locations to all be unique modulo M . If this is true, then at any time T , the set $(K + T) \bmod NM$ will only have one element that is zero modulo M , and therefore only one collimator hole overlaps with a source spot and only one source fires.

Volumetric coverage from a source having multiple rows and a 2D detector can now easily be achieved by simply assigning slots to different source rows. For R source rows, we use NRM snapshots and we partition them into the R source rows. Specifically, if the j th of the NRM snapshots is $r \bmod R$, then the source spot will be energized on the r th source row. In this manner, the sampling pattern within a single source row is effectively repeated and rotated by a fraction of a view in each successive source row.

To minimize aliasing artifacts, we can choose Φ so that the samples from multiple snapshots of the small detector perfectly interlace, analogous to detector quarter offset or flying focal spot. One subtlety here is that the conversion from Φ in radians to K in slots may induce rounding error. In our design below, we chose Φ to be uniformly spaced and selected DID so that the rounding error would be zero.

II.F. Virtual bowtie

Variations on this basic timing scheme allow the system to become more flux efficient. By redistributing the firing sequence, the firing density could be higher in some parts of the object projection than in other parts, as opposed to a uniform distribution of firing density throughout. As an example, for smaller objects such as the head, it may be possible to halve the FOV to 27 cm instead of 54 cm (Table III). Instead of energizing a source location at the collimator holes where the photons pass outside the field of view, Φ could be chosen so that firing sequence is distributed uniformly in a smaller range. This would require additional collimator holes to be placed closer to the center of the collimator arc. In other cross-sections, we might choose Φ such that at any given time there are multiple collimator holes that are zero modulo M , providing the system with the option to shoot from any one of multiple source locations. These timing schemes can implement a *virtual bowtie*,¹¹ so that the system can allocate photons on an as-needed basis and tailor the radiation profile to the imaging task and thereby improve the dose efficiency. Previous work has estimated that the use of a virtual bowtie could improve dose efficiency by a factor of two without increasing the peak variance of the reconstruction.^{11,33}

To estimate the flux advantage of the virtual bowtie, let us consider an idealized system with a perfect virtual bowtie that is able to send an arbitrary number of photons along each ray in the sinogram. To simplify the discussion, assume a parallel-beam scan of a uniform, centered water cylinder of radius R and linear attenuation coefficient μ . The total field of view of the system is a circle with radius F . While these assumptions are simplistic, they allow for a straightforward analytic estimate of the flux advantage.

Suppose a reference system samples each ray in the sinogram with I_0 photons. Because the water cylinder to be scanned is radially symmetric, it suffices to consider a single view. The number of photons detected as a function of the detector cell location from the center channel, which we will denote as $D(x)$, is

$$D(x) = \begin{cases} I_0 \exp(-2\mu\sqrt{R^2 - x^2}), & |x| \leq R \\ I_0, & |x| > R \end{cases}.$$

The virtual bowtie could be used to equalize the transmitted flux, providing uniform image quality across the patient. Physical bowties are designed for this objective but do not perfectly achieve it because they can only be designed for generic patient shapes. More importantly, a physical bowtie can reduce the incident intensity at x by attenuating the beam but cannot increase the intensity at $x = 0$. The virtual bowtie, by contrast, reduces the intensity at x by reducing the time during which the sources fire toward x and therefore can dedicate more time, and therefore more flux, at $x = 0$. For a given ray, a distance x from the isocenter, the flux advantage of the virtual bowtie is $D(x)/D(0)$. On average, the flux advantage of the perfect virtual bowtie over the physical bowtie across the field of view is therefore

$$\frac{1}{2F} \int_{-F}^F \frac{D(x)}{D(0)} dx.$$

For operating conditions of $\mu = 0.2 \text{ cm}^{-1}$, $R = 15 \text{ cm}$, and $F = 25 \text{ cm}$, this flux advantage is 3.5. However, we note that even for a maximally sized patient with $R = 25 \text{ cm}$, the virtual bowtie still produces a flux advantage of 2.6. The actual implementation of virtual bowtie in SS-IGCT will include discretization limitations and will not be perfect, though, nonetheless, elliptical or thoracic cross-sections are better scanned with a dynamic, virtual bowtie than a static, physical bowtie. Overall, we believe that a flux advantage of 3 is reasonable for most general-purpose scans, and that a flux advantage of 5 should be achievable for targeted scans. In a targeted scan, high SNR would be achieved in only a small region of interest, such as a single organ. Alternatively, in the targeted scan a truncated dataset could be collected and the reconstruction could be performed with recently developed algorithms.³⁴ Conventional systems could also implement targeted scanning with appropriate beam shaping filters to improve their dose efficiency, but these filters cannot improve the effective flux.

II.G. Flux

Although the proposed system has the flux advantage of the virtual bowtie, available flux remains a concern for this system. The shorter minimum rotation time, which is three times faster than most systems today, and the small area of the detector present challenges for flux. In spite of this, we believe that the flux of this system could be sufficient for many clinical applications.

The fundamental instantaneous power limit of x-ray sources is thermal. Conventional CT sources employ the “line focus” principle and a rotating anode to spread out the thermal impact.¹ However, even with rotation, a point on the anode may be illuminated for tens of microseconds during each anode rotation. By contrast, the focal spot dwell time for the current scanned anode (SBDX) source is only $1 \mu\text{s}$, and the anode surface area in the proposed scanned source is very large. Even with targeted scans and the virtual bowtie, with the x-ray source concentrated on an area of source several cm in length, the time before any given spot is re-energized is on the order of milliseconds. This is several times longer than the current SBDX system and long enough to allow the target to cool almost completely between exposures. If necessary, additional source rows could be used to further increase the cooling time. These factors combine to offset the fact that the current SBDX does not use line focus, so that normalized for effective focal spot size, the current SBDX is quite comparable in power to x-ray sources in clinical CT scanners (100 kW/mm^2). The scanned anode could realize a further power increase by a reduction in source spot dwell time. We estimate that reducing the dwell time to $0.5 \mu\text{s}$ would allow a tube power increase of 75%. The transmission anode design of the SBDX system may be 70% more efficient than conventional reflection anodes and additionally produce an optical spot which is similar across the field of view.³⁵ Alternatively, a reflection anode with line focus may be used to increase the instantaneous power. We estimate that the power of the x-ray source could increase fivefold before reaching instantaneous thermal limits on the anode. In this regime, the maximum tube power may be constrained by other considerations, but overall we believe that much higher power sources are possible.

Fast, wide cone-beam scanners with 16 cm of volumetric coverage are well suited for particular clinical applications, including whole-organ perfusion, cardiac function testing, and coronary artery imaging. Because whole-organ perfusion requires tens of repeated scans of the same organ, typical protocols call for relatively low power. At our clinic, brain and liver perfusion studies are routinely conducted at 100 mAs per scan, about a third of the flux limit of a modern CT system with a rotation time of 0.3 s. Cardiac function tests are conducted at even lower power. Coronary artery imaging uses the full power of the scanner ($\sim 1000 \text{ mA}$), but because of cardiac motion, a half-scan is typically used so that the total technique is about 200 mAs.

Table I compares the flux available from the proposed system to that of a wide cone-beam system with a 16 cm of coverage. The shorter exposure time and smaller detector of the SS-IGCT system together reduce the available flux by

TABLE I. System flux comparison to a wide-cone reference scanner for imaging tasks requiring wide, volumetric scanning. SS-IGCT may produce flux sufficient for many clinical applications, such as whole-organ perfusion.

Factor	Reference scanner, 16 cm	SS-IGCT	Relative flux (%)
Exposure time	300 ms	100 ms	33
Detector size	100 × 32 cm	10 × 16 cm	5
Bowtie filter	Physical	Virtual, targeted	500
Detector	Energy integrating, antiscatter grid	Photon counting, no antiscatter	140
Anode	Rotating reflection	Transmission or large target reflection	300
Reconstruction	FBP	Iterative	150
Total			53

a factor of 60 compared to the peak available from a conventional system. As discussed previously, we estimate the virtual bowtie recovers a factor of 5 for targeted scans. An optimized scanned source can recover another factor of 3 using a transmission anode, or up to a factor of 5 using a reflection anode. To be more conservative, we assumed a factor of 3. The use of a photon-counting detector may increase the dose and flux efficiency of the scan by reducing Swank noise and electronic noise. In addition, the elimination of the antiscatter grid, made practical by the reduction in detector area size, increases the geometric efficiency. We estimate this combined effect to be at least a factor of 1.4. Finally, the emerging use of iterative reconstruction improves image quality at reduced dose and flux.³⁶ Although estimates of the benefit of iterative reconstruction vary widely, we assumed that comparable image quality can be achieved with two-thirds of the flux. We see that the net effect is to bring the SS-IGCT system into rough parity with the conventional wide cone-beam system. Although the temporal resolution is much higher and the detector area is much smaller, the proposed system is still able to produce sufficient flux for applications such as whole-organ perfusion or cardiac function testing. Of course, the conventional system also could and should adopt iterative reconstruction, photon-counting detectors, etc., and reduce its power requirements, but the purpose of our flux comparison is to show that the majority of scans which are possible with current systems may still be achievable with SS-IGCT.

Most routine clinical scans do not demand extremely high temporal resolution. In these applications, scanners with 4 cm of axial coverage per rotation have proven to be sufficient.

Table II compares the flux of the proposed system with a clinical helical scanner with 4 cm of axial coverage per rotation for these typical imaging tasks. We assumed the reference clinical scanner rotates in 350 ms and operates with a helical pitch of 1 to sample a 16 cm volume. For this comparison, since the ultimate temporal resolution is not required, we allowed the SS-IGCT to use a single axial scan of 350 ms to capture the entire volume. We see again that the flux between the two systems remains comparable, and by imaging the entire volume in a single scan, SS-IGCT avoids artifacts, which may be associated with helical scans and motion in between rotations. If desired, SS-IGCT may also produce targeted, follow-up images with greater temporal resolution or better noise characteristics.

Some factors are not listed in Tables I and II but still impact effective flux. In SS-IGCT, the distribution of flux throughout the volume is not uniform, although the number and spacing of source rows could be modulated to reduce this effect. However, we note that for the reference scanner, modern helical reconstruction algorithms are not always flux-efficient because the data may be reweighted to reduce helical artifacts at the expense of increasing noise.³⁷ In spectral imaging tasks, the use of an energy-discriminating, photon-counting detector could provide higher contrast-to-noise ratio, reducing the flux requirements.

It is clear that SS-IGCT is not the architecture of choice for high-dose imaging tasks because the small detector size intrinsic to IGCT places a large penalty on flux. Conventional systems are better suited for applications that require very high x-ray intensities. In addition, if even greater flux is desired in conventional systems, improvements in source

TABLE II. System flux comparison to a standard clinical scanner operating in helical mode to capture a 16 cm volume. SS-IGCT captures the volume in a single axial scan and with increased temporal resolution.

Factor	Reference scanner, 4 cm	SS-IGCT	Relative flux (%)
Exposure time	1400 ms	350 ms	25
Detector size	100 × 8 cm	10 × 16 cm	20
Bowtie filter	Physical	Virtual, general	300
Detector	Energy integrating, antiscatter grid	Photon counting, no antiscatter	140
Anode type	Rotating reflection	Transmission or large target reflection	300
Reconstruction	FBP	Iterative	150
Total			95

technology such as better cooling or the use of multiple target tracks may further improve the power output of these systems. Use of multiple sources and detectors multiplies flux for conventional systems (albeit at the expense of increased cost and scatter). However, the comparisons to conventional systems above show that an optimized SS-IGCT could produce sufficient flux for many clinical applications, and, therefore, flux does not prevent feasibility.

II.H. Other system design issues

The parameters for our system are presented in Table III. The number of source rows used directly impacts the level of cone-beam artifacts. The Tuy criterion states that accurate reconstruction is not possible if any plane passing through the volume to be reconstructed does not intersect with the source trajectory.³⁸ The Tuy criterion is closely related to Tuy's inversion formula, which can be thought of as using plane integrals to measure the frequency content of an object. It is impossible to meet the Tuy criterion with any finite number of parallel source rings because most plane integrals parallel to the source rings cannot be measured. The frequency data corresponding to these missing plane integrals are therefore missing. However, currently used clinical systems operating in axial mode do not meet the Tuy criterion but produce acceptable images because the fraction of frequency content missing is small. As the cone angle of the system increases, the fraction of frequency data missing also increases, producing more severe cone-beam artifacts. In our choice of parameters, successive source rings are placed 4 cm apart, making the fraction of missing plane integrals comparable to existing scanners with 4 cm of axial coverage operating in step-and-shoot mode, which produce images that are routinely judged to be acceptable. Previous work demonstrating cone-beam artifacts for reconstruction algorithms with multiple parallel axial rings is consistent with these expectations.^{25,39} Figure 5 shows the sampling in a longitudinal plane.

It would be preferable to increase the sampling in the longitudinal plane further, but limitations on the source and on

TABLE III. Proposed system parameters. The “worst case minimum cone angle” refers to the smallest tilt angle for which any voxel along isocenter will be seen by at least one source row. This is diagrammed in Fig. 5.

Source-to-isocenter distance	60.0 cm
Detector-to-isocenter distance	50.3 cm
Inplane field of view (diameter)	54 cm for circular objects
Axial or z-field of view	16 cm
Maximum cone angle	8.28°
Detector size	10 (transverse) × 16 cm (axial)
Number of detector cells	100 × 160
Detector pixel size	1 × 1 mm
Source focal spot size	0.5 mm
Number of source arrays	3
Source array length	120 cm
Gaps between source array	6 cm
Number of views (N) including missing views	1200
Missing views (in gap)	63
Firings per view (M)	21
Number of source rows	5
Worst case minimum tilt angle	1.91°
Spacing between source rows	4 cm
Total number of source firings	126 000
Source spot dwell time	750 ns
Move time between source locations (masked)	250 ns
Total scan time	95 ms

the detector frame rate do not permit an arbitrarily dense sampling of the source. The sampling characteristics of the system are therefore constrained by hardware limitations. A faster source and detector would allow a reduction of the source spot dwell time, enabling higher instantaneous power and more rows. If a slower source and detector are necessary, this can be accommodated by using fewer source rows, increasing cone-beam artifacts, or fewer views per source row, increasing view sampling artifacts, although we note that the angular locations of sources on different rows could be interleaved to reduce these view sampling artifacts.

FIG. 5. Sampling in a longitudinal (for example, a coronal) plane with the proposed system parameters. (a) The SS-IGCT system, with five source rows and with maximum cone angle α . (b) A reference, conventional CT system with 4 cm of coverage in the axial direction. Its worst case minimum tilt angle (equivalently, the half-cone angle) is α .

III. SIMULATIONS

We simulated the proposed system to ensure that the reconstruction did not degrade the resolution or noise characteristics and to ensure that the system is in fact capable of suppressing cone-beam artifacts. All of our results are compared to a reference system, which is a simple cone-beam axial scan with the same axial coverage and without any gaps using the standard FDK algorithm for reconstruction. The reference system has the same focal spot size and detector pixel size as the proposed system but no missing rays, and it is expected to have cone-beam artifacts. All simulations were monochromatic.

Simulations were conducted using analytic projection models to estimate the initial CT data. During the reconstruction process, a simple, voxelized forward projection model was used.

To evaluate the spatial resolution, a noiseless simulation of a 100 μm diameter sphere at isocenter was performed. Only the center three of the five source rows illuminate the central slice, so only these source rows were used in the reconstruction process. An ideal ramp filter was used in the reconstruction. A large number of sourcelets and detectorlets were used to simulate the finite size source spots and detector cells.¹ To decrease the computational time necessary, the number of views used was decreased, but we did not expect this to change the image quality in a targeted field of view.

To ensure that the noise properties were not degraded by the system geometry and reconstruction algorithm, the noise was measured using a 3 cm water sphere at the isocenter. The expected number of photons at each detector cell in the absence of any attenuation was 10^6 photons. The resulting statistics are then assumed to be Poisson distributed. Here again, only the center three source rows and a reduced number of views were used. The total number of photons available to both systems was kept constant.

To examine possible artifacts we simulated a phantom based on the FORBILD thorax phantom.⁴⁰ The vertebrae of the thorax phantom are known to produce strong cone-beam artifacts at larger cone angles. A Hanning window was applied for both the reference system and the SS-IGCT system in these reconstructions.

IV. RESULTS

Figure 6 compares the MTF in the inplane and axial directions of the proposed system to the reference FDK system without gaps. With both systems the inplane resolution is superior to the axial resolution because of inplane oversampling and Fourier interpolation. In the axial direction, linear interpolation is instead used during backprojection. The MTF of the proposed system is slightly worse than the reference system, perhaps due to the use of some forward projected rays.

Figure 7 shows the noise in one slice of the 3-cm diameter water sphere. The standard deviation in a region of interest inside the water sphere was 320 HU for the reference system and 301 HU for the proposed system. The small reduction in the noise is consistent with the slight degradation of the

FIG. 6. Modulation transfer function of the proposed system as compared to the FDK reference.

resolution. Because the number of photons available to both the reference FDK system and the SS-IGCT system are the same and because the distribution of photons is uniform in angle for both systems, the total dose delivered is equivalent. We conclude that the photon utilization efficiency of the two systems is therefore quite similar.

With the SS-IGCT system, Fig. 7(b) shows some correlated noise. This comes from higher noise streaks in the direction of the forward projected rays. These forward projected rays are produced from the starting reconstruction, which is noisy because only rays with small tilt angles are used, and these rays are not resolution matched to the measured rays. If these issues were corrected, some fundamental noise discrepancy would still exist because of the nonuniform sampling of the views.

Figure 8 shows images of the thorax phantom. Compared to the reference system, the proposed system has drastically reduced cone-beam artifacts, which is apparent both in the sagittal and axial slices. The proposed system shows slight streak artifacts along rays that connect sharp edges to gaps. We believe these are due to the simplicity of the forward projector we used which can cause the resolution

FIG. 7. Noise performance of (a) the reference FDK system and (b) our proposed system. Window width is 800 and level is 0. The standard deviations in the water spheres were measured to be 320 HU for the reference system, and 301 HU for our proposed system.

FIG. 8. Sagittal slices of the (a) reference system and (b) the proposed system, and axial slices through the (c) reference system and (d) the proposed system for a FORBILD thorax phantom. Window width is 200 and level is 0. [Reprinted with permission from S. S. Hsieh and N. J. Pelc, "A volumetric reconstruction algorithm for stationary source inverse-geometry CT," *Proc. SPIE 8313, Medical Imaging 2012: Physics of Medical Imaging*, 83133N (2012)].

of the forward-projected rays to be different from measured rays, and discontinuities in projections when the transitions between measured and missing rays is aligned with sharp edges. The inconsistency at the transition can create streaks. More sophisticated forward projectors may alleviate these gap artifacts.

V. CONCLUSIONS

We have detailed a system combining stationary source arrays with the IGCT concept, thereby achieving very fast temporal resolution in conjunction with reduced scatter, reduced cone-beam artifacts, and increased dose efficiency through the virtual bowtie. Three separated source arrays are more feasible to build than a complete ring but produce gaps in the sinogram. A multistep reconstruction scheme is adopted which minimizes the effects of these gaps and produces reconstructions of sufficient quality. In simulations of the thorax phantom, the artifacts induced by the gaps are much smaller in magnitude than the cone-beam artifacts that are suppressed by the use of multiple source rows. Solutions for other potential challenges, including collimation and timing, are proposed. The flux of the system is a challenge but should be sufficient for most clinical applications. As presented, the system is capable of achieving a complete volumetric scan with 16 cm of axial coverage within 100 ms with spatial resolution com-

parable to conventional FDK cone-beam systems, therefore, meeting our performance targets. The design can be scaled to even greater volumetric coverage while still keeping cone-beam artifacts manageable. Compared to previous IGCT designs, SS-IGCT has better temporal resolution. Whereas previous IGCT designs used a single, complex rotating source array, SS-IGCT uses a series of large and stationary source arrays.

Although this initial study suggests that the SS-IGCT system is capable of being constructed and produces reconstructions with adequate image quality, more work is required to fully explore its advantages. While it may be possible to build a system with three separated source arrays, a system with nine source arrays may be more directly actionable, requiring simpler modifications from existing scanned source array platforms such as the SBDX system.^{19,35} The algorithms discussed here will not work with nine gaps, however, so specific algorithms must be developed to meet these requirements. Iterative techniques promise the flexibility of working with this unusual dataset, but developing an analytic reconstruction technique would be of interest as well. Several analytic reconstruction algorithms have been proposed in recent years to deal with unusual source trajectories,^{22–24} but with a few exceptions,²⁸ most assume that the trajectory is continuous. Stationary source array systems require very different assumptions and the source trajectory can be extremely

discontinuous, so research is needed to adapt existing algorithms to this new problem. It may also be feasible to use multiple detectors¹² in conjunction with SS-IGCT, allowing for even faster temporal resolution or for greater flux. Finally, besides theoretical discussions and simulation, it will be essential to collect experimental data to better explore the possibilities of this potential system architecture. Although more work remains to be done, SS-IGCT appears to be a promising design, which combines fast temporal resolution with very large volumetric coverage and excellent dose efficiency.

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