Direct two-dimensional reconstruction algorithm for an inverse-geometry CT system

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An inverse-geometry computed tomography (IGCT) system uses a large source array opposite a smaller detector array. A previously described IGCT reconstruction algorithm uses gridding, but this gridding step produces blurring in the reconstructed image. In this article, the authors describe a two-dimensional (2D) IGCT reconstruction algorithm without gridding. In the transverse direction, the raw data of the IGCT system can be viewed as being composed of many fan beams. Because the spacing between source spots is larger than the spot width, each fan beam has undersampled projection data, but the missing samples are effectively provided by other undersampled fan beam views. In the proposed method, a direct fan beam reconstruction algorithm is used to process each undersampled fan beam. Initial images with this method showed ring artifacts caused by nonuniform sampling in the radial direction as compared to an ideal fan beam. A new method for correcting this effect was developed. With this correction, high quality images were obtained. The noise performance of the proposed 2D IGCT reconstruction algorithm was investigated, and it was comparable to that of the fan beam system. A MTF study showed that the proposed method achieves better resolution than the gridding method. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3050160](https://doi.org/10.1118/1.3050160)]

Key words: computed tomography (CT), inverse-geometry CT, MTF, noise, gain correction

I. INTRODUCTION

Important factors in computed tomography (CT) imaging include the desire for faster scan times, reduced motion artifacts, improved dose efficiency, and higher resolution. Inverse geometry CT (IGCT) has been proposed¹ as a technique that may offer an improved compromise among all these factors. As shown in Fig. 1, in a full 3D implementation, it uses a large-area 2D array of x-ray sources and a 2D detector array with a smaller extent in the transverse direction. IGCT enables accurate reconstruction of a thick 3D volume in a single rotation without the cone beam artifacts caused by the data insufficiency problems of cone beam CT with a single circular orbit of the x-ray source.² IGCT also allows flexible control of the spatial distribution of incident x-rays (“virtual bowtie”) that can lead to a substantial reduction in x-ray dose for the same peak image noise level.³ In addition, a smaller detector area may provide efficient scatter rejection and a cost advantage compared to the conventional cone-beam system.

A reconstruction algorithm for the IGCT system, developed by Schmidt *et al.*^{4,5} rebins the data into parallel rays by using a gridding method. However, the gridding step can introduce blurring in the reconstructed image. This article presents a direct IGCT reconstruction algorithm that does not require gridding. Because the transverse direction data sets acquired by the IGCT system are very similar to a group of fan beams, we sought to develop an algorithm that relies on the fan beam reconstruction algorithm⁶ avoiding the approximation inherent in gridding.

The key aspects of the algorithm can be fully appreciated

in the 2D reconstruction context. Therefore, a 2D algorithm will be presented and tested and then the generalization to 3D is discussed. We first provide a brief description of the IGCT geometry and the proposed reconstruction algorithm. We then illustrate the difference in the ray sampling patterns in 2D Radon space between the IGCT system and a fan beam system, which is caused by the different geometry. It is shown that the different radial sampling patterns can cause ring artifacts in the reconstructed image, and a new method is presented to correct it. The quantum noise performance of the proposed reconstruction algorithm is investigated analytically and by computer simulation, and compared with that of a fan beam system. Finally, a MTF study is performed to compare the resolution of the proposed method to that of the gridding method.

II. METHOD

II.A. System geometry

The simple 2D IGCT system we use to describe and test the direct reconstruction method is shown in Fig. 2. The source array consists of 40 focal spots, each collimated to illuminate the detector array. The source array is wider than the detector array which is composed of 80 elements. For the current study, an offset detector array is used to avoid data redundancy. The specifications of the system are summarized in Table I. As can be seen in Fig. 2, each x-ray focal spot generates small fan beam projection data by illuminating the entire detector array. The source and detector mounted on a gantry rotate around the patient during the data acquisition. A “superview” is the collection of rays from all sources to all

FIG. 1. The IGCT system with a large source array and a small detector array.

detector elements at a single gantry position. Since the rays span a wide range of angles in 2D Radon space, only 278 superviews are needed for our field of view (FOV) and are equivalent to a fully sampled fan beam system with 1112 views, from a view sampling perspective.

II.B. Direct reconstruction method

At each gantry angle, the IGCT data can be viewed as many small fan beams, each composed of projection rays connecting one detector element with all sources. In our example with 80 detector elements and 40 x-ray focal spots in each row, each superview has 80 detector-based fan beam views. Each of these fan beams is at a slightly different angle and a different radial alignment to the center of rotation. In addition, each fan beam is undersampled in the radial direction because the source spot spacing is much larger than the source spot size (i.e., 5 and 0.5 mm, respectively, in the system of Table I). However, the other small fan beam views from the other detectors provide additional rays at different radial distances from the center of rotation. As illustrated in Fig. 3(b), with enough suitably placed detector elements, sufficient sampling is obtained and high quality imaging should be achievable.

In the IGCT geometry, each ray from a source location to a detector location is described by two parameters, the

FIG. 2. Transverse view of the IGCT system.

TABLE I. Specifications for the IGCT system.

Source-to-center-detector distance	1000 mm
Source-to-isocenter distance	438.5 mm
Source array width	200 mm
Detector array width	25.6 mm, one sided
Source array spot spacing	5 mm
Detector array element spacing	0.32 mm
Source spot width	0.5 mm
FOV (transverse)	120 mm
Number of superviews	278/360°

signed distance between the ray and the isocenter, ρ , and the angle between the ray and the line from the source center to the isocenter, θ . These two parameters are described in Fig. 4 and can be calculated as follows:

$$\theta = 90 - \text{atan}\left(\frac{Sy - Dy}{SDD}\right), \quad (1)$$

$$\rho = \frac{SID}{SDD} Sy \times \sin \theta, \quad (2)$$

where SDD is the source to detector distance, SID is the source to isocenter distance, and Sy and Dy are the signed distances of the source and detector elements, respectively, from the central ray (e.g., the line that contains the center of the source array and the center of rotation). These two parameters specify one sample point in 2D Radon space.

In the IGCT system of Table I, the source spacing is ten times larger than the source spot width. According to the Shannon sampling criterion and assuming the sampling aperture is spot width limited, this ratio has to be 0.5 to avoid aliasing. Thus, each small fan beam is undersampled by as much as a factor of 20. A conventional fan beam system uses focal spot wobbling⁷ or a detector quarter-offset to achieve the Shannon sampling rate.⁸ As discussed above, in the IGCT system, the additional ρ samples to achieve the Shannon sampling criterion come from other detector elements.

We propose to use a method based on an equally spaced fan-beam filtered backprojection for image reconstruction. Each undersampled projection will be filtered and back-projected. To avoid blurring in the filtering step, we insert 19 zeros between true source positions and assume the missing samples in one zero-inserted fan beam are compensated by

FIG. 3. Projection geometry of (a) a standard fan beam and (b) the IGCT system for a simple case of two detector elements. Dotted lines represents missing rays in the undersampled IGCT fan beams.

FIG. 4. Description of two geometry parameters for a ray in the IGCT geometry.

other zero-inserted fan beams. We used a Hanning windowed reconstruction filter. The final image was produced by back-projecting the filtered zero-inserted fan beams, taking care to preserve the geometry of each.

There is likely to be a reconstruction time penalty for the proposed method due to the large number of filtering and backprojection steps, one per detector and superview. In the current example, the data consists of $80 \times 278 = 22\,240$ undersampled views which are equivalent to a fully sampled fan beam system with 1112 views. Thus, we have an order of magnitude more filtering and backprojection steps. This is partly offset by the elimination of the gridding step.

II.C. Simulation

To evaluate the reconstruction algorithm, an analytical Shepp–Logan phantom was simulated. “Step and shoot” projection data were produced by calculating the line integrals along each ray through the phantom. The source and detector apertures were simulated by averaging projection data from discrete sub-sources and sub-detectors spanning the finite aperture sizes. Figure 5 shows the reconstruction of the Shepp–Logan phantom using the method as just described. Prominent ring artifacts can be seen. As will be shown below, these artifacts are due to non-uniform sampling in the ρ direction.

II.D. ρ sampling error and correction

The ρ values sampled by the ensemble of rays of the IGCT system are not exactly the same as those in a conventional fan beam. In the simple example of Fig. 3, even though the conventional and IGCT systems each has eight rays at eight ρ values, the ρ values are not exactly same because of the different geometries. The specific ρ values sampled by an IGCT system depend on the number and spacing of sources and detectors and on the other system parameters. The sampling density in the ρ direction therefore can be nonuniform and not smooth. When gridding is used, the nonuniform sampling is taken into account by the density correction. In the proposed algorithm, the nonuniform sampling is equivalent to nonuniform gain as a function of ρ . As

(a)

(b)

FIG. 5. (a) Reconstructed image and (b) central profile using the direct reconstruction method.

shown in Fig. 5, this causes ring artifacts in the reconstructed image.

In filtered-backprojection reconstruction with a conventional fan beam, cosine weighting factors are used to correct for the smoothly varying and analytically predictable ρ sampling density. We propose to correct for the nonuniform ρ sampling density in the IGCT system by multiplying the projection data by geometry-specific gain or weighting factors. Because the errors are geometry dependent and not object specific, the same gain factors should work for all objects.

The goal, therefore, is to find a gain vector such that when it is multiplied by the measured projection and reconstructed, an image comparable to the 2D fan beam reconstruction is produced. To find the gain vector, the reconstruction steps, filtering, and backprojection, are expressed as a matrix operation. For a projection with m samples (including zeros

inserted between real samples), m gain values and $2m-1$ reconstruction filter coefficients are used. A centered-uniform cylinder which is large enough for the gain vector estimation was used so that the projection is the same for all views. We will then verify that the same gain vector works for other

objects. Note that the object used to calculate the gain vector must be as large as the largest object to be imaged.

The gain multiplication and filtering for each zero-inserted fan beam can be conducted by the following matrix multiplication:

$$\begin{aligned}
 F_i = H(P_i \cdot G_i) &= \begin{bmatrix} h_1 & 0 & \cdots & 0 & 0 \\ h_2 & h_1 & \cdots & \vdots & \vdots \\ h_3 & h_2 & \cdots & 0 & 0 \\ \vdots & h_3 & \cdots & h_1 & 0 \\ h_{2m-2} & \vdots & \cdots & h_2 & h_1 \\ h_{2m-1} & h_{2m-2} & \vdots & \vdots & h_2 \\ 0 & h_{2m-1} & \cdots & h_{2m-3} & \vdots \\ 0 & 0 & \cdots & h_{2m-2} & h_{2m-3} \\ \vdots & \vdots & \vdots & h_{2m-1} & h_{2m-2} \\ 0 & 0 & 0 & \cdots & h_{2m-1} \end{bmatrix} \begin{bmatrix} p_{1,i}g_{1,i} \\ p_{2,i}g_{2,i} \\ p_{3,i}g_{3,i} \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ p_{m,i}g_{m,i} \end{bmatrix} \\
 &= \begin{bmatrix} h_1p_{1,i} & 0 & \cdots & 0 & 0 \\ h_2p_{1,i} & h_1p_{2,i} & \cdots & \vdots & \vdots \\ h_3p_{1,i} & h_2p_{2,i} & \cdots & 0 & 0 \\ \vdots & h_3p_{2,i} & \cdots & h_1p_{m-1,i} & 0 \\ h_{2m-2}p_{1,i} & \vdots & \cdots & h_2p_{m-1,i} & h_1p_{m,i} \\ h_{2m-1}p_{1,i} & h_{2m-2}p_{2,i} & \vdots & \vdots & h_2p_{m,i} \\ 0 & h_{2m-1}p_{2,i} & \cdots & h_{2m-3}p_{m-1,i} & \vdots \\ 0 & 0 & \cdots & h_{2m-2}p_{m-1,i} & h_{2m-3}p_{m,i} \\ \vdots & \vdots & \vdots & h_{2m-1}p_{m-1,i} & h_{2m-2}p_{m,i} \\ 0 & 0 & 0 & \cdots & h_{2m-1}p_{m,i} \end{bmatrix} \begin{bmatrix} g_{1,i} \\ g_{2,i} \\ g_{3,i} \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ \cdot \\ g_{m,i} \end{bmatrix} = A_i G_i, \quad (3)
 \end{aligned}$$

where F_i is the filtered projection for the i th zero-inserted fan beam, P_i is the projection data for the i th zero-inserted fan beam, G_i is the $m \times 1$ gain vector for the i th zero-inserted fan beam, and H is a $(3m-2) \times m$ matrix of filter coefficients. A_i is a $(3m-2) \times m$ matrix whose terms contain filter coefficients and the i th zero-inserted fan beam data.

The final image is produced by summing the backprojections of each zero-inserted fan beam. Pixel-driven backprojection of each zero-inserted filtered projection using linear interpolation can be described by the following equation. (Note that for the case of a centered uniform cylinder, A_i is independent of superview angle.)

$$R_{\text{IGCT}} = \sum_{i=1}^{N_d} \left(\sum_{j=1}^{N_v} L_{ij} A_i \right) G_i, \quad (4)$$

where N_d is the number of detectors, N_v is the number of supervIEWS, L_{ij} contains the linear interpolation coefficients

of the i th zero-inserted fan beam at the j th view, and R_{IGCT} is the reconstructed image.

In Eq. (4), L_{ij} and A_i are known for any known object, and G_i is unknown. By comparing R_{IGCT} with the reconstruction from an ideal fan beam, R_{FAN} , the error vector E is defined as

$$E = R_{\text{IGCT}} - R_{\text{FAN}}. \quad (5)$$

The nonzero components of the gain vector which minimizes the L_2 norm of E was found by using a conjugate-gradient method,⁹ using $\mathbf{1}^T$ as the initial guess. The root-mean-square (rms) error was calculated as in each iteration

$$\text{rms}_E = \frac{\|E\|_2}{\sqrt{n}}, \quad (6)$$

where n is the length of the error vector E . For a particular IGCT sampling geometry, once the gain vector is known, a reconstruction of arbitrary data is performed by multiplying the gain vector by the IGCT projection data set prior to filtered backprojection.

II.E. Noise performance

In a fan beam system, the variance, σ^2 , at a pixel can be expressed as.^{6,10}

$$\sigma^2 = \left(\frac{2\pi\tau}{M} \right)^2 \sum_{i=1}^M \frac{w_i^2}{U_i^4 N_i} \sum_k h_k^2, \quad (7)$$

where M is the number of views, N_i is the number of detected photons that passed through the pixel at the i th view, w_i is the cosine weighting factor at the i th view, h_k is the filter coefficients, U_i is the magnification factor from the reconstructed point to the detector at the i th view, and τ is the sampling distance between projection data.

In direct IGCT reconstruction, the cosine weighting factor w_i is replaced by the gain value g_i . Therefore, the variance equation of the IGCT image can be expressed as follows:

$$\sigma^2 = \left(\frac{2\pi\tau}{M'} \right)^2 \sum_{i=1}^{M'} \frac{g_i^2}{U_i^4 N_i} \sum_k h_k^2, \quad (8)$$

where M' is the number of rays that passed through the pixel over all superviews.

Equations (7) and (8) were used to analytically compare the noise performance of a fan beam system with that of the IGCT system with our direct reconstruction. Noise performance was also studied using computer simulation. In the simulation, a virtual bowtie was assumed so that quantum noise for a detected signal of 50 photons per measurement, uniform over all rays, was used. The value of the sampling distance τ was 0.014 cm. 200 noise realizations were simulated and the noise variance and standard error in the variance were measured by using a central 3 mm by 3 mm region of interest. In order to probe the noise performance across the FOV, we measured a variance map as a function of radial distance for the IGCT system and for a conventional fan beam system, and then their variance ratio was calculated. For comparison, the variance map of a gridding method with two different gridding kernels for the IGCT system was also examined.

TABLE II. Parameters for rebining.

1D projection sampling distance	0.014 cm
Number of rebinned views over 2π	1112
Angular kernel width	0.0113 rad
Radial kernel width	0.028 cm
Minimum sum of corrected kernel values	0.1
Maximum δ value	1000

II.F. MTF

The goal of the direct reconstruction algorithm is to avoid blurring caused by gridding. To compare the resolution performance of these methods, a 0.0035 cm radius sphere at isocenter and also one at (3 cm, 3 cm) were simulated. Images were reconstructed using the proposed direct IGCT reconstruction algorithm and an IGCT reconstruction using gridding and the parameters in Table II. As a reference, parallel beam data for the same objects and the geometry of Table II were simulated and reconstructed with direct filtered backprojection. MTF curves were calculated by taking 2D Fourier transform of the reconstructed images. To fully explore the resolution loss due to gridding, the IGCT data were also rebinned using kernels that were two and three times wider than those in Table II in both the angular and radial directions. To examine the resolution performance of the proposed direct method as a function of radial position, the MTF was calculated at four different locations (0.1 mm, 0), (1.8 cm, 0), (3.6 cm, 0), and (5.5 cm, 0). For all MTF studies, the same source and detector blurring were simulated, and the Hanning cutoff of 40 lp/cm was used.

III. RESULTS

III.A. Gain vector calculation

The modeled IGCT system has 3200 rays per superview, and therefore needs a gain vector composed of 3200 values. Figure 6(a) shows the rms error per iteration. The initial rms

FIG. 6. (a) Rms error per iteration and (b) gain vector for the IGCT geometry of Table I.

FIG. 7. (a) Reconstructed image and (c) central profile of the IGCT system with using a gain vector. (b) Reconstructed image and (d) central profile of the IGCT system with a gridding method. Reconstructed noisy images of (e) a direct method and (f) a gridding method.

error was 0.2598, and after 1000 iterations it was reduced to 0.0019. Figure 6(b) shows the gain vector calculated using the data of a uniform centered cylinder after 1000 iterations.

The Shepp–Logan phantom data were reconstructed by the proposed method, and a gridding method. Figure 7(a) shows the reconstructed image with the gain vector correc-

tion, and Figure 7(c) plots the central profile. Comparison with Figs. 5(a) and 5(b) shows that the ring artifacts were suppressed significantly. This result supports the approach of calculating a gain vector for a uniform centered cylinder and applying it to an arbitrary object. Figures 7(b) and 7(d) show the reconstructed image and a central profile using a gridding

TABLE III. Variance comparison.

	Theoretical noise variance	Measured noise variance $\pm$ standard error
Fan beam system	0.00676	$0.0073 \pm 5.5 \times 10^{-5}$
IGCT system	0.00690	$0.0078 \pm 7.7 \times 10^{-5}$

method which employs a kernel that is two times wider than those in Table II in both the angular and radial directions. Noisy projections of the Shepp–Logan phantom were also reconstructed. Figures 7(e) and 7(f) show the reconstructed noisy images using the direct method and a gridding method. The gridding interpolation degrades the resolution but also reduces image noise, as demonstrated further below.

III.B. Noise performance

With Eqs. (7) and (8), the theoretical noise variance was calculated for a fan beam system and the IGCT system. Since the IGCT system with 278 superviews is comparable to a fan beam system with 1112 views, $M=M'=1112$ were used in Eqs. (7) and (8). Table III shows the theoretical variance as well as the measured variance and standard error in the variance for a central $3\text{ mm} \times 3\text{ mm}$ region of interest, and Fig. 8 shows the ratio of the variance of IGCT with the direct method or with gridding to that of a fan beam system across the FOV. These results show that the noise performance of the IGCT system with direct reconstruction is comparable to that of the fan beam system, and the gridding method shows different noise performance depending on the kernel size. Noise reduction with increased radial distance is caused by blurring in the projection angular direction.

III.C. MTF

Figures 9(a) and 9(b) compare the MTF of the direct IGCT reconstruction, the IGCT reconstruction with gridding,

FIG. 8. Measured ratio of variance of a fan beam system to that of the IGCT system as a function of radial position.

FIG. 9. (a) MTF of a direct method, a gridding method, and a parallel beam system at center. (b) MTF of a direct method, a gridding method, and a parallel beam system at off-center. (c) MTF of a direct method at four different locations.

and the parallel beam reconstruction at two locations. As can be seen, the direct reconstruction algorithm provides a better resolution than a gridding method and achieves a resolution comparable to that of a parallel beam system. Figure 9(c) shows that the resolution performance of the proposed

method is consistent across the FOV. The MTF degradation with the gridding method is caused by the convolution of the Radon space data with the gridding kernel and gets worse with increasing gridding kernel width. The resolution loss is worse for the off-center object due to blurring in the azimuthal image direction caused by the gridding interpolation in the angular direction. To minimize resolution loss, smaller gridding kernel widths should be used. However, with a smaller gridding kernel width more data points have to be acquired to ensure that all output points are adequately covered. Figure 9(b) demonstrates that the MTF of IGCT with gridding approaches the MTF of the parallel beam system when a half-width gridding kernel and twice as many projections are used. By contrast, the MTF of the direct reconstruction matches that of the parallel beam system with the nominal number of projections.

IV. DISCUSSION AND CONCLUSIONS

We have shown that it is possible to reconstruct 2D IGCT data using a variant of the direct fan beam filtered backprojection method. The IGCT data can have nonuniform radial sampling that depends on the particular system geometry. Even slight nonuniformity of the sampling density can cause ring artifacts with the proposed method, but these can be prevented by applying a multiplicative gain factor prior to filtering and backprojection. A new method to compute the gain correction was shown to be effective. Noise studies showed that noise penalty due to the gain vector is negligible while the MTF simulations demonstrated that the proposed algorithm achieves a spatial resolution that is better than a gridding method and comparable with that of a direct parallel beam system. The advantage of the proposed algorithm is that it avoids gridding and its associated resolution loss, while the disadvantage is a large reconstruction time penalty due to the need for more convolution and backprojection steps.

Ultimately, one may desire to extend the method to a 3D implementation. When the source and detector arrays have the same extent in the axial direction, the IGCT data include approximately “in-plane” rays (rays from one source row to a detector row at approximately the same axial location) for each slice. The proposed method can be directly used for in-plane data at each location to provide accurate reconstructions. If the sampling pattern (i.e., relative source and detector element locations) is the same for each slice, the same gain vector can be used at each location. In general, the IGCT system also provides “cross-plane” rays which connect a source row to a detector row at a different axial location,

and these can be used to improve SNR. In order to use all the rays for the direct 3D IGCT reconstruction, we need a generalization of the direct 2D fan beam reconstruction. One approach that could be used would be based on 3D FDK. Gain vectors would be calculated by simulating a centered-uniform cylinder and again solving for the gains that minimize the rms difference between the IGCT reconstruction and that with uniformly sampled data. Variants of 3D filtered backprojection^{11,12} may also be feasible. If the source and detector locations are the same in each row, we expect the gain vectors to be the same for all cross plane angles, and this could be used to reduce the size of the problem being solved. A full investigation of the extension of our method to 3D is beyond the scope of this article. The described method for computing the gain correction can be used for any CT system that suffers from unknown gain factors and for which the reconstruction can be written as a single matrix. This includes all x-ray CT and PET scanners.

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