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A Multi-Source Inverse-Geometry CT system: Initial results with an 8 spot x-ray source array

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Abstract

We present initial experimental results of a rotating-gantry multi-source inverse-geometry CT (MS-IGCT) system. The MS-IGCT system was built with a single module of 2×4 x-ray sources and a 2D detector array. It produced a 75 mm in-plane field-of-view (FOV) with 160 mm axial coverage in a single gantry rotation. To evaluate system performance, a 2.5 inch diameter uniform PMMA cylinder phantom, a 200 μm diameter tungsten wire, and a euthanized rat were scanned. Each scan acquired 125 views per source and the gantry rotation time was 1 second per revolution. Geometric calibration was performed using a bead phantom. The scanning parameters were 80 kVp, 125 mA, and 5.4 μs pulse per source location per view. A data normalization technique was applied to the acquired projection data, and beam hardening and spectral nonlinearities of each detector channel were corrected. For image reconstruction, the projection data of each source row were rebinned into a full cone beam data set, and the FDK algorithm was used. The reconstructed volumes from upper and lower source rows shared an overlap volume which was combined in image space. The images of the uniform PMMA cylinder phantom showed good uniformity and no apparent artefacts. The measured in-plane MTF showed 13 lp/cm at 10% cutoff, in good agreement with expectations. The rat data were also reconstructed reliably. The initial experimental results from this rotating-gantry MS-IGCT system demonstrated its ability to image a complex anatomical object without any significant image artefacts and to achieve high image resolution and large axial coverage in a single gantry rotation.

Keywords

CT; Inverse geometry; multiple source array; image quality

1. INTRODUCTION

Over the past few decades, the axial coverage of multi-slice 3rd generation CT has been continually increased, enabling imaging of thicker volumes with fast scan times, thin slices, and reduced motion artefacts in a single gantry rotation [Fishman et al 2001, Kohl G 2005, Schuijf et al 2009]. While these technical developments enable innovative clinical applications such as CT angiography, perfusion imaging, and cardiac imaging, the larger detector size and cone angle increased detected scatter radiation, cone-beam artefacts, heel effect, and detector cost.

To overcome several limitations in wide cone angle CT, a new concept, multi-source inverse-geometry CT (MS-IGCT), was proposed [De Man et al 2007a, De Man et al 2007b, De Man et al 2009, Schmidt et al 2004]. While conventional multi-slice CT scanners use a point x-ray source and a large detector array, the MS-IGCT system has a distributed source array and a relatively small detector array as shown in figure 1. Each source in the transverse direction emits a narrow x-ray beam through a portion of the field-of-view (FOV) which enables the system to achieve uniform spatial resolution across FOV, and additional sources in the longitudinal direction increase the volumetric coverage and reduce cone-beam artefacts. Because of its small detector size, the scatter and detector cost can be reduced. In addition, the x-ray flux of each source can be modulated independently, ('virtual bowtie') which can minimize the effective dose to the patient by optimizing the x-ray flux of each source [Sperl et al 2009].

Recently, a gantry-based MS-IGCT research prototype was implemented and several experiments were performed. The purpose of this paper is to present initial experimental results of this rotating-gantry MS-IGCT system and demonstrate the performance of the system in studies of phantoms and a complex anatomical object.

2. MATERIALS AND METHODS

2.1 Sampling, spatial resolution, and noise considerations

Since the MS-IGCT system has somewhat different characteristics than a conventional 3rd generation multi-slice CT system, we discuss here some of the similarities and differences with respect to sampling, spatial resolution, and noise.

Each focal spot of the MS-IGCT system illuminates a fraction of the horizontal field of view and generates Radon space data for a range of radial distances from isocenter. The design strategy that has the minimum number of sources is one in which the radial distances from the set of sources abut each other and barely cover the desired field of view. If the system has more sources that are closer together, it will generate overlapped data in the radial direction in Radon space. The overlapped data improve sampling in Radon space and ensure that the system does not suffer from aliasing and potentially increase spatial resolution in the same way as focal spot deflection or quarter detector offset [Hsieh J 2003]. The overlapped data also improve the measurement statistics. If the number of views and total scan time are held constant, as the number of sources is increased the time allocated to each source (and possibly the x-ray pulse width) must decrease and the detector frame rate must increase. A

shorter x-ray pulse width can be helpful in enabling higher x-ray tube power. However, the higher detector frame rate may be challenging and may lead to higher electronic noise.

As in a conventional CT system, the in-plane MTF in MS-IGCT is given by [Hsieh J 2003]:

$$MTF(f)_{system} = MTF(f)_f \times MTF(f)_d \times MTF(f)_a \times MTF(f)_r \quad (1)$$

where $MTF(f)_f$, $MTF(f)_d$, $MTF(f)_a$, $MTF(f)_r$ represent the MTF of focal spot size, detector response, azimuthal blurring, and reconstruction algorithm, respectively. Even though the geometry of the MS-IGCT system is “inverted”, the x-ray magnification is exactly as it is in a conventional x-ray system, SDD/SOD, where SDD is the source-to-detector distance and SOD is the source-to-object distance. For any ray, the impact of the focal spot size, the detector aperture, and the magnification are exactly as in any other x-ray system. In conventional systems, the apparent focal spot varies across the field of view [Hsieh J 2003]. It is significantly wider at fan angle positions comparable to or larger than the x-ray tube target angle and causes a degradation of spatial resolution in the radial direction with increasing distance from isocenter. This effect is less severe in MS-IGCT since the fan angle from each source is less than in a conventional system. Azimuthal blurring depends on the ratio of the effective exposure time per sample to the rotation time, and also increases with increasing radial distance [Hsieh J 2003]. The exposure time per measurement in an MS-IGCT system is necessarily very short because of the need to sample data from each of the sources. With these short x-ray pulses the effect of azimuthal blurring (i.e., $MTF(f)_a$) is negligible in MS-IGCT. The reconstruction algorithm MTF is as in any CT system, with the rebinning step introducing some blurring as is the case in conventional systems that use this approach. Thus, the main difference between MS-IGCT and conventional systems in this regard are in azimuthal blurring and focal spot effects, with the MS-IGCT system expected to have more uniform resolution throughout the FOV. The dependence of resolution in the slice direction is the same in the two designs.

As discussed in detail below, images are produced by rebinning to cone-beam data sets followed by reconstruction using an FDK algorithm [Feldkamp et al 1984], and the raw data set can have overlapped sampling depending on the number of sources and the size of the detector array. The rebinning step combines multiple raw data samples. The statistical noise in each rebinned measurement therefore depends on the number of photons in each raw measurement and the degree of overlapped sampling. After rebinning, the noise propagates into the image in a manner similar to a conventional system [Hsieh J 2003, Baek et al 2011]. The effective mAs depends on the x-ray tube current and the time that each object region is illuminated. The latter depends on the x-ray pulse width and the number of pulses that illuminate a particular region, which is generally the number of measured multi-source views times the number of overlapped samples.

2.2 Tested system overview

The tested MS-IGCT system employed a single source module and a 2 dimensional detector array [Frutschy et al 2009, Frutschy et al 2010, Neculaes et al 2010, Uribe et al 2010]. The detector uses modules from the GE CT750HD system because of its speed and improved low-signal performance. Each detector module consists of 64×16 cells of about $1.1 \text{ mm} \times$

1.0 mm each, and 12 detector modules are tiled in z producing a total sensitive area of about 70 mm × 192 mm. Figure 2 shows the full detector assembly. The source module (shown in figure 3) has two rows of four focal spots separated in z by 100 mm, which enable 160 mm axial (z) coverage. The focal spot spacing in each row is 25 mm, and therefore the in-plane FOV of this single module system is about 75 mm diameter. If three more source modules were added, the in-plane FOV would be about 220 mm diameter. The maximum tube voltage and current were 140 kVp and 250 mA for these experiments, and the gantry rotation speed is one revolution per second.

Figure 3 shows the source module, with two rows each having four focal spots/cathodes, the anode assembly between these two rows, and a vacuum chamber with the source module. Each row consists of a 4×1 electron gun – an electron gun that has four identical but independently controlled electron emitters. Electron emitters used here are dispenser cathodes; each emitter is circular, with a diameter of 3.5 mm. Reliability, lifetime and good emission properties are the reasons why commercial grade dispenser cathodes have been chosen for this application. All four emitters in the 4×1 electron gun have the same beam extraction and focusing scheme. The beam extraction is performed using a mesh grid, made from molybdenum, located at about 300 μm from the electron emitting surface. The common grid plate has four mesh grid circular patches, located on top of each of the emitters; the power electronics control scheme allows individual control for the extraction voltage for each emitter. The mesh grid has a transparency of about 20 % - this means that 80 % of the cathode emitting current reaches the target, while ~ 20 % will be collected by the mesh grid. Typical grid voltages are on the order of few hundreds of volts, to get emission levels of hundreds of mA. The source is typically operated in pulse mode, with pulse widths on the order of tens up to 100 μs. All eight cathodes in the source presented in figure 3 are kept at about 1100 C, but emission is triggered only when a positive voltage is applied on the cathode surface. Typical pressure in the vacuum chamber is on the order of 10⁻⁷ Pa. The 4×1 electron gun includes two electrodes for beam optics: ECE (emittance compensating electrode) and focusing electrode (figure 3). These voltages are typically DC and all four electron beams in a 4×1 gun see the same voltages for beam optics control. The anode includes a copper block with cooling channels, and tungsten targets brazed onto the copper block (figure 3).

The MS-IGCT system was built starting from a GE Lightspeed VCT gantry. Many of the VCT components were used, including the rotating gantry, the high-voltage generator, the heat exchanger, and the power distribution unit. Detector modules, readout electronics and slip-ring were borrowed from GE CT750HD scanner. We designed and manufactured brackets to hold all the new components, including the new source and detector at the designed g-forces, including a high safety factor. Figure 4 shows the integrated MS-IGCT gantry.

For data acquisition, the source spots were energized sequentially, and each spot illuminated a small portion of the entire FOV. Each complete acquisition consisted of 125 views per source equally spaced over 360° and adjacent sources shares an overlap region in 2D Radon space [Baek et al 2012a], which provided sufficient view sampling for the current in-plane

FOV. Data were acquired from phantoms and a post-mortem rat. For all experiments, the system operated at 80 kVp and 125 mA. Table 1 summarizes the system parameters.

2.3 Geometric calibration

Accurate geometric calibration is an important step in producing high quality CT images. In the MS-IGCT system, the axis of rotation may not be aligned with the z-axis, and source locations may not be positioned at the intended positions. For calibration, we assumed that the detector coordinates are 9 known, while the source locations and the axis of rotation were unknown. We characterized each source location by 3 parameters (i.e., x_s , y_s , and z_s), and the axis of rotation by 4 parameters (i.e. x_offset , y_offset , α and β), as depicted in figure 5 (a). The goal of the geometric calibration is to estimate these unknown parameters, using a scan of spherical beads.

To estimate the unknown parameters, we first formulated the forward problem, that is, determining the line integral of beads given the source locations, bead information, and axis of rotation. Then, the calibration problem is reduced to solving the inverse problem using a non-linear least squares algorithm. In the inverse problem, bead information such as exact bead locations, bead radius, and the attenuation coefficient of the bead were treated as unknowns.

A calibration phantom was designed using a 4-cm diameter empty plastic cylinder with five beads placed on the surface, and separated by 2 cm along the axis of the cylinder (shown in figure 5 (b)). Projection data were acquired at 125 view angles equally spaced over 360 degrees. Since the projection images contained effects of the cylinder as well as the beads (shown in figure 5 (c)), bead-only projection images were estimated by subtracting the cylinder projection data in rows near a bead from the projection data through the bead. From the bead-only projection images, bead centers were found using a matched filter, and a threshold was applied to extract the bead projection. The extracted bead projection data were used to solve the inverse problem using a non-linear least-squares algorithm. While a simulation study with the same unknown parameters and realistic noise showed good accuracy with an estimation error less than tens of microns, effects such as beam hardening and scattered photons present in real data may affect the estimation performance. To reduce the estimation error, a 200 μm diameter tungsten wire phantom was scanned and the estimated parameters were iteratively refined to maximize the reconstructed value of the wire.

2.4 Data preprocessing and reconstruction

During the data acquisition, the intensity of each x-ray source can fluctuate, for example, due to small variations in the output of the power supply. This can produce image artefacts [Baek et al 2012a]. While conventional multi-slice CT scanners monitor source intensity fluctuations using a ‘reference channel’ at the end of the detector array (outside the FOV), several sources in the MS-IGCT system cannot see the ‘reference channel’ since each source illuminates a small portion of the entire FOV which makes it impossible to illuminate a reference channel. Correction for the intensity fluctuations was performed by using a data normalization technique [Pelc et al 2010]. In addition, beam hardening and spectral

nonlinearity of the detector were corrected by using a second order polynomial fit, where the fitting parameters of each detector channel were estimated by comparing the analytic projection data of a uniform cylinder phantom with the scan data of the uniform PMMA cylinder phantom [Hsieh J 2003].

For image reconstruction, the preprocessed projection data were rebinned into a full fan beam data set for each detector row, and therefore two full cone beam data sets, one each from the upper and lower source rows. Figures 6 (a) and (b) show a transverse view of data acquisition geometry and the sampling pattern of the MS-IGCT system in 2D (parallel ray) Radon space. In the rebinning step, we searched input grid points within a rectangular fit region centered at the output grid point as shown in figure 6 (b), and estimated the value of output grid point using a first order 2D function:

$$f(\rho, \theta) = a\rho + b\theta + c, \quad (2)$$

where ρ is radial distance, θ is the projection angle, and a , b , and c are the coefficients of the first order 2D function.

To capture enough input grid points within the fit region, the rebinning fit widths were set to 0.9 mm and $\sim 1^\circ$ along the radial and angular directions, respectively. Since the noise levels of input grid points were different due to the non-uniform illumination from each x-ray source as shown in figure 7, the coefficients of the 2D function were estimated by using a weighted least squares method in which the detected intensities of input grid points were used as the weighting values [Baek et al 2008]:

$$\hat{\beta} = (X^T W X)^{-1} X^T y, \quad (3)$$

where X is a 2D array of input grid point locations, y is a column vector of projection data of input grid points, W is a 2D array of detected intensities of input grid points, and $\hat{\beta}$ is a vector with the estimated coefficients (a , b , and c) of the first order 2D function in Equation (1). As a result, the output grid point estimated from the fitted 2D function produced the optimal SNR. The projection data were rebinned into 250 views since the adjacent sources provide additional samples (i.e., the sampling redundancy is a factor of two with the system of table 1).

The rebinned full cone beam data sets were reconstructed using an FDK algorithm [Feldkamp et al 1984], where the reconstruction filter was apodized by a Hanning window with a cutoff of 17 lp/cm. The reconstruction parameters are summarized in table 2. The reconstructed voxel size was determined by considering the redundancy (~ 2) and magnification (~ 2.1). The volume reconstructed from each source row was about $7.5 \times 7.5 \times 10.5 \text{ cm}^3$. Because of larger detector size in the z direction, reconstructed volumes from upper and lower source rows shared an overlap volume of $7.5 \times 7.5 \times 5 \text{ cm}^3$. The two volume reconstructions in this overlap volume were blended in image space by applying smoothly varying weighting function [Baek et al 2012b].

2.5 Phantom and post-mortem animal studies

The uniformity of the reconstructed CT values and image artefacts were examined by using reconstructed coronal and axial planes and the central profile of the axial plane of a 2.5 inch diameter uniform PMMA cylinder phantom. To measure the in-plane MTF of the MS-IGCT system, a 200 μm diameter tungsten wire aligned parallel and close to the axis of rotation was scanned, and a volume of $3.75 \times 3.75 \times 3.5 \text{ cm}^3$ centered at isocenter was reconstructed with a voxel size of $0.15 \times 0.15 \times 0.14 \text{ mm}^3$ and a Hanning window reconstruction filter with a cutoff of 34 lp/cm. To minimize the estimation error caused by image noise, a point spread function (PSF) in the central axial slice of the reconstructed volume was fitted with a Gaussian function [Nickoloff et al 1985], and then the in-plane MTF was estimated by taking FFT of the fitted Gaussian. We also simulated a 200 μm diameter wire at the isocenter, and calculated the in-plane MTF. The estimated in-plane MTF was compared with the simulated in-plane MTF. Since the current system covers 7.5 cm FOV, we expected the variation of the spatial resolution across the FOV is very small, and therefore we only measured the system MTF near the isocenter. To demonstrate the system performance for more complex, anatomical objects, a post-mortem rat was also scanned.

3. RESULTS

Figures 8 (a) and (b) show coronal and axial images of the reconstructed uniform PMMA cylinder phantom. The central 5 cm region of the coronal image is less noisy than the other regions due to the blending of the overlap volume, but the reconstructed images do not show any image artefacts. Figure 8 (c) shows the central profile of the axial image (i.e., figure 8 (b)), where the uniformity of the reconstructed values can be observed. Figures 8 (d) - (f) show coronal, sagittal, and volume rendered images of the rat. These images demonstrate that the complex object can be reconstructed reliably by the MS-IGCT system. Figure 9 compares the simulated and measured in-plane MTF curves; both show 10% modulation at 13 lp/cm and are in good agreement throughout. These initial results with a single source module (i.e., 2×4 x-ray sources) demonstrate the feasibility of MS-IGCT imaging, and its potential for yielding large volumetric coverage with good image quality. This also shows the feasibility of the proposed distributed source architecture.

4. CONCLUSIONS

We describe the MS-IGCT system integration and present initial testing results using a rotating-gantry, the first demonstration of this type of imaging architecture. The imaging results demonstrated that the system can image a complex anatomical object without any significant image artefacts and achieve high image resolution and large axial coverage in a single gantry rotation.

In this study, the spatial resolution was investigated near the isocenter since the current system has a relatively small in-plane FOV. In a conventional CT system, the optical focal spot width varies as a function of fan-angle, which, combined with azimuthal blurring, produces nonuniform spatial resolution across the FOV [Kyprianou et al 2005, Kyprianou et al 2006]. Because each source in the MS-IGCT system only illuminates a small fan-angle and the x-ray pulse widths are short, much more uniform spatial resolution can be expected

across the FOV. We could not explore this effect with this small field of view system; it will be verified experimentally with the 32-spot x-ray source, the final embodiment of the distributed x-ray source for MS-IGCT.

The MS-IGCT system does not employ an anti-scatter grid, and therefore the full detector area can be utilized. This helps to improve the detective quantum efficiency (DQE) and potentially reduce the dose. In a conventional multi-slice CT system, the absence of the anti-scatter grid would increase the scatter-to-primary ratio (SPR) significantly. By contrast, the MS-IGCT system has relatively low SPR because each source illuminates a small fraction of the entire volume. We found that the amount of scatter in the small FOV system was very small, and therefore scatter correction was not applied. Even for a 300 mm diameter water phantom and for a 100 mm $\times$ 100 mm detector, Monte Carlo simulations have shown that the maximum SPR is only about 15% [De Man et al 2007].

In a recent study, De Man et al. investigated the potential benefits of the ‘virtual bowtie’ (control of intensity as a function of time for each source) on image noise and dose, and showed that, at the same effective dose, the peak image variance was reduced by 61 % compared to that in a conventional multi-slice CT system employing a bowtie filter and using an optimal mA modulation [De Man et al 2007b]. This result demonstrates the potential benefits of the MS-IGCT system for dose reduction. In our source implementation, the beam current from each of the sources cannot be rapidly modulated but the pulse width is fully programmable. This feature has been tested but not exploited in imaging studies yet. In the current 8 spot system, only small objects can be imaged so the range of detected x-ray flux is small. We plan to investigate the ‘virtual bowtie’ concept experimentally after adding three more source modules.

Our current system has several limitations, especially on source power. The limit of the dispenser cathodes is about 1500 mA, certainly not a bottleneck in the current multispot x-ray source. HV stability may limit the operation of the tube. In initial tests, repeated “spits” occurred and set the limit on the power we could achieve so far. Continued seasoning experiments can eliminate spits and we expect active cooling would reduce the frequency of spits. The most fundamental limiting factor is the target temperature. The source power is ultimately limited by the melting temperature of tungsten. Assuming a cold target, 120 kVp tube voltage, a 1 mm $\times$ 1 mm optical focal spot ($\sim 6 \text{ mm}^2$ thermal area) and a 100 μs dwell time, the maximum tube current in a single x-ray pulse would be about 1000 mA. However, for a 1 s scan with hundreds of pulses per spot and a repeat time of 43 ms the temperature in the focal spots and surrounding area will build up, and the maximum tube current is only 250 mA. For even longer scan times, the bulk temperature of the source starts to rise and oil cooling becomes critical. Oil cooling is part of the design but has not yet implemented. For comparison, clinical scanners have tube currents of at least 500 mA for this small focal spot size and rotation speeds of 0.35 s. Our system has a rotation speed of 1 s and a voxel duty cycle (fraction of time any given voxel is irradiated) of about 5%. Hence, the integrated flux would be 14 times lower with our research experiments. The impact on image noise would be partly offset by the increased DQE and the advantage of the dynamic bowtie. Experimentally we can further compensate for this by performing slower scans. Eventually, one could develop the MS-IGCT scanner with reduced source-to-isocenter distance,

improved detector efficiency, reduced dwell times, increased detector size, ‘virtual bowtie’ and statistical reconstruction [Beque et al 2007] to achieve acceptable clinical noise levels.

In summary, we described system integration and presented initial testing results of the first rotating-gantry MS-IGCT system. The imaging results demonstrated that the current system can image a complex anatomical object without any significant image artefacts and achieve high image resolution and large axial coverage in a single gantry rotation.

Acknowledgments

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Figure 1.
The system geometry of (a) conventional multi-slice CT and (b) MS-IGCT.

Figure 2.
Detector using detector modules of a GE CT750HD scanner.

Figure 3.

(a) The X-ray source module with two rows of four focal spots/cathodes each, (b) side view of the 4×1 electron gun and target, and (c) a vacuum chamber with one source module.

Figure 4.
Picture of the MS-IGCT gantry in rotation.

Figure 5.

(a) Diagram of the system for the geometric calibration, (b) bead phantom, and (c) sample projection image and extracted bead-only projection image.

Figure 6.

(a) The transverse view of data acquisition geometry and (b) corresponding sampling pattern of the MS-IGCT system in 2D Radon space. In Figure 6 (b), solid line represents the sampled projections of each source, and the dotted line represents the rebinned projection.

Figure 7. Exemplified detected intensity of each source. Non-uniform illumination of each source is clearly seen.

Figure 8. Reconstructed images from scan data. (a) Coronal image, (b) axial image, and (c) central profile of the axial image from the reconstructed uniform PMMA cylinder phantom. (d) Coronal image, (e) sagittal image, and (f) volume rendered image of the rat. The units of the reconstructed values is cm^{-1} , and the display window of the reconstructed uniform PMMA cylinder phantom and a rat are [0.15 0.27] and [0.15 0.35].

Figure 9.
Comparison of the measured and simulated in-plane MTF curves.

Table 1

System parameters.

Source to isocenter distance	450 mm
Detector to isocenter distance	385 mm
Source spacing in x and z	25 mm, and 100 mm
Detector dimension in x and z	70 mm $\times$ 192 mm (64 $\times$ 192)
Detector cell size in x and z	1.1 mm $\times$ 1.0 mm
FOV(axial $\times$ transverse)	75 mm $\times$ 160 mm
Source voltage	80kVp
Source current	125 mA
Exposure time per source location per view	5.4 μ s
Number of views per source	125/360°
Rotation time per revolution	1s

Table 2

Reconstruction parameters.

Rebinned view	250/360°
Rebinning fit width, radial direction	0.9 mm
Rebinning fit width, angular direction	~1°
Reconstructed voxel size	$0.3 \times 0.3 \times 0.28 \text{ mm}^3$
Reconstructed volume size of each source row	$7.5 \times 7.5 \times 10.5 \text{ cm}^3$ (250 × 250 × 375)
Reconstruction filter cut off	17 lp/cm