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Preliminary Feasibility of Dedicated Breast CT With an Inverse Geometry

Taly Gilat-Schmidt
Marquette University, tal.gilat-schmidt@marquette.edu

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Taly Gilat Schmidt

[a]Department of Biomedical Engineering, Marquette University, Milwaukee, WI, USA;

ABSTRACT

In this study we theoretically investigated the minimum scan time of an inverse-geometry dedicated breast CT system that provides sufficient sampling and dose equivalent to mammography without exceeding the limits of source power or detector count rate. The inverse geometry, which utilizes a large-area scanned source and a narrower photon-counting detector, is expected to have improved dose efficiency compared to cone-beam methods due to reduced scatter effects and improved detector efficiency. The analysis assumed the specifications of available inverse-geometry source and detector hardware (SBDX, NovaRay, Inc, Newark CA). The scan time was calculated for a 10, 14, and 18-cm diameter breast composed of 50% glandular / 50% adipose tissue. The results demonstrate a minimum scan time of 6.5, 14.3, and 14.7 seconds for a 10, 14, and 18-cm-diameter breast, respectively. The scan times are comparable to those of proposed cone-beam systems. For all three breast sizes, the scan time was limited by the detector count rate. For example, for the 14-cm-diameter breast, the minimum scan time that met the source power limitation was 1.1 seconds, and the minimum scan time that achieved sufficient sampling was 0.8 seconds. The scan time can be reduced by increasing the detector count rate or area. Effective bowtie filters will be required to prevent detector saturation at the object edges. Overall, the results support preliminary feasibility of dedicated breast CT with an inverse geometry.

Keywords: Breast CT, Inverse geometry, Photon counting, Breast imaging

1. INTRODUCTION

Dedicated breast CT systems have been proposed to provide volumetric imaging of the breast without breast compression and with dose equivalent to mammography.[1,2] Breast imaging is a challenging task that requires low radiation dose, high spatial resolution to depict calcifications, and high contrast resolution to depict masses. Most proposed dedicated breast CT systems are based on the cone-beam geometry, illustrated in Figure 1. The patient lies prone, with the breast to be imaged placed in the pendant position through an opening in the table. The CT gantry is located below the table, consisting of an x-ray source emitting a half-cone of x-rays towards a large-area x-ray detector. Most proposed breast CT systems employ a conventional x-ray tube and an energy-integrating flat-panel detector.

The cone-beam geometry has several fundamental image quality limitations. One limitation is that a circularly-scanned cone-beam acquisition does not provide sufficient sampling, which can lead to cone-beam artifacts.[3] A second limitation is the high scatter that results when a large volume is irradiated. Detected scatter reduces the contrast-to-noise ratio (CNR) and introduces artifacts.[4] The scatter effects of a dedicated breast CT system were experimentally measured, and the study found the scatter-to-primary ratio (SPR) varied from 0.2 to 1.0 depending on breast size and composition.[5] A multi-slit multi-scan breast CT system was proposed with a photon counting detector and multiple slot apparatus to reduce scatter.[6] The system is based on the cone-beam geometry and requires mechanical scanning of the collimators to scan the field of view.

The inverse geometry was proposed to provide volumetric CT imaging while overcoming the limitations of cone-beam CT.[7,8] Inverse-geometry CT (IGCT) uses a large-area scanned source and a photon-counting detector array that is narrower in the transverse direction, as illustrated in Figure 2. While imaging, the electron beam is electromagnetically steered across the transmission target of the source, dwelling behind each of an array of collimator holes which limits the x-ray beam to illuminate the detector array. For each source position, the

Further author information: (Send correspondence to T.G.S.)
taly.gilat-schmidt@marquette.edu

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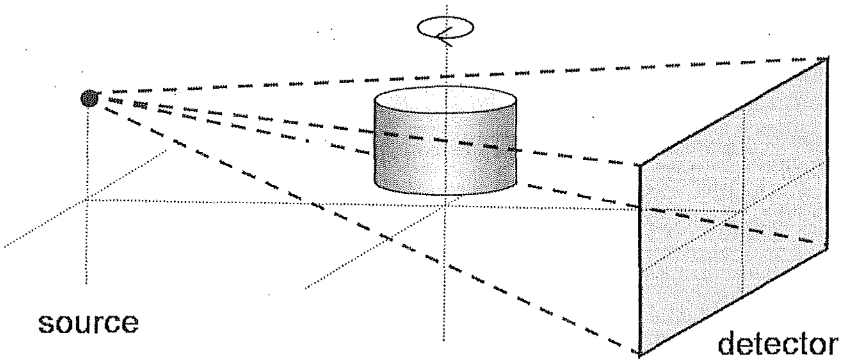

Figure 1. Cone-beam dedicated breast CT geometry consisting of an x-ray source and a large-area detector.

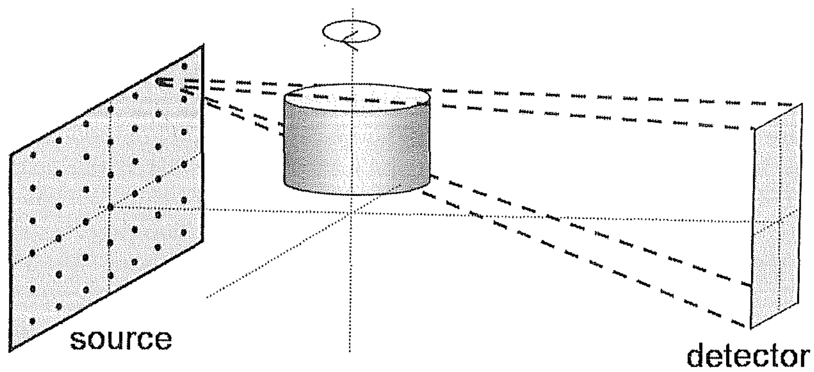

Figure 2. Inverse-geometry dedicated breast CT consisting of a large-area scanned source and a narrower detector array.

entire detector array is read, forming a truncated projection image. Sufficient sampling and negligible cone-beam artifacts are achieved because the source and detector have the same axial extent. The detected scatter is reduced because only a fraction of the object is irradiated at any time and because the photon-counting detector is fast enough to isolate measurements from successive source locations. A simulation study found an order of magnitude reduction in SPR for the inverse geometry compared to the cone-beam geometry for dedicated breast CT.[9] Due to this reduction in scatter, the inverse-geometry improved the CNR by a factor of 1.2-1.4. Additional dose efficiency is expected due to the improved detective quantum efficiency of the photon-counting detector.[10]

The proposed IGCT system relies on innovative hardware components including a scanned transmission x-ray source and a photon-counting detector. While these novel components enable the benefits of IGCT, these components also pose technical challenges that must be addressed in order to demonstrate feasibility for dedicated breast scanning. During an IGCT acquisition, the x-ray beam emitted at each source location irradiates a fraction of the field of view for a fraction of a second. The beam is off while the source is steered to the next location. Due to this scanning procedure, every voxel in the scanned volume is irradiated for a fraction of the total scan time. In contrast, a cone-beam acquisition continuously illuminates all voxels in the field of view during a scan. This short scan time per voxel raises the question of whether the IGCT source can provide sufficient photon fluence within a reasonable scan time. Assuming the source can produce the required flux, there is the additional question of whether the detector is fast enough to count all incoming photons.

In this study we ask the following question: what is the minimum scan time in which a dedicated breast IGCT system can image a range of breast sizes at dose equivalent to mammography with sufficient sampling and without overloading the source or saturating the detector?

Table 1. Inverse-geometry dedicated breast CT specifications

Source Extent (in-plane x slice)	40 x 18 cm
Source Sampling (in-plane x slice)	160 x 72 positions
Detector Extent (in-plane x slice)	5.4 x 18 cm
Detector Sampling (in-plane x slice)	48 x 160 pixels
Source-to-Isocenter Distance (SID)	31.5 cm
Detector-to-Isocenter Distance (DID)	46.5 cm
Field of View (in-plane x slice)	22 x 18 cm

2. METHODS

2.1 System Specifications

The first step in this study was to specify an IGCT configuration suitable for dedicated breast CT. The specifications, listed in Table 1, were determined by inverting a cone-beam dedicated breast CT geometry studied in the literature.[5] The hardware specifications were based on the source and detector components of the Scanning-Beam Digital X-ray (SBDX) system (NovaRay, Inc. Newark, CA), modified to match the field of view required for breast CT.[11,12] The SBDX transmission target is comprised of a water-cooled thin-film layer of tungsten-rhenium on a beryllium substrate. The detector is a photon-counting direct-conversion detector that is a tiled array of detector hybrids, each hybrid consisting of a CdTe tile bumpbonded to a custom silicon IC. The transmission source dwells for one microsecond at each position in the array, and spends 0.28 microseconds moving between positions. The SBDX source has a maximum power of 24 kW, therefore, at 80 kVp the maximum tube current is 300 mA. The maximum detector count-rate is approximately 5.5 million counts per second.

We next considered the constraints of source power, detector count rate, and sampling for a 10, 14, and 18-cm-diameter breast composed of 50% glandular / 50% adipose tissue imaged at 80 kVp with dose equivalent to mammography.

2.2 Source Considerations

Every voxel in an IGCT acquisition is irradiated for a fraction of the total scan time: the time the voxel is in the field of view of the x-ray beam and the time that the beam is on. If we consider the voxel at isocenter, the fraction of time that the voxel is irradiated during an acquisition (f_{time}) is equal to[7]

$$f_{time} = \left(\frac{Area_{det}}{Area_{source}} \cdot \frac{SID^2}{DID^2} \right) \cdot f_{duty} \tag{1}$$

where $Area_{det}$ is the total detector area, $Area_{source}$ is the area of the source array, SID is the source-to-isocenter distance, DID is the detector-to-isocenter distance, and f_{duty} is the fraction of time the beam is on. For the SBDX source, f_{duty} is equal to 1/1.28.

Combating this fractional irradiation time are several factors that increase the IGCT source flux at isocenter per mA compared to a cone-beam system. In most proposed dedicated cone-beam breast CT systems, the DID is smaller than the SID. When a conventional cone-beam geometry is inverted, isocenter moves closer to the source than in a comparable cone-beam geometry thereby increasing the flux at isocenter. Also, the transmission anode has an estimated 1.7 times greater photon flux per mA than the reflection anodes used in conventional x-ray imaging.[11] Combining these two effects, the factor (f_{flux}) which describes the increased IGCT photon flux per mA at isocenter is

$$f_{flux} = \left(\frac{DID^2}{SID^2} \right) \cdot 1.7 \tag{2}$$

The ultimate question in this study is how quickly the source can provide the needed photon fluence without exceeding the power limits. To answer this question, the required photon fluence must be determined. Boone et. al., have calculated technique settings for their cone-beam breast CT system.[13] If we assume that the 80kVp spectrum shape is the same for both the IGCT system and that used in Boone's study, the minimum IGCT scan time that provides the same photon fluence at isocenter is

$$T_{source} = \frac{mAs_{cone}}{f_{time} \cdot f_{flux} \cdot mA_{max}} \tag{3}$$

where mAs_{cone} is listed in the published technique tables for each breast size and mA_{max} is the maximum tube current available at 80 kVp (300 mA for the SBDX source). Eq. 3 expresses that in order to match the mAs of a cone-beam system, and assuming a fixed IGCT tube current, the IGCT scan time must be adjusted to account for the fraction of time each voxel is scanned and for the increased IGCT photon flux per mA.

2.3 Detector Considerations

Assuming the source can produce the required flux, an additional question is whether the detector is fast enough to count all incoming photons. Boone et. al., have calculated the raw-beam photon fluence at isocenter that results in dose equivalent to mammography for a range of breast sizes. Assuming a raw-beam photon fluence (photons/mm^2) of F at isocenter, the number of photons that pass through a voxel at isocenter and reach one detector pixel over the entirety of the scan in the absence of an object (N_{raw}) is equal to

$$N_{raw} = \frac{F \cdot Area_{pixel} \cdot SID^2}{(SID + DID)^2} \tag{4}$$

where $Area_{pixel}$ is the area of one IGCT detector pixel, and F is the photon fluence at isocenter that provides dose equivalent to mammography for the 10, 14, and 18-cm-diameter breast.[13]

Assuming the breast has attenuation μ and diameter d, and that the maximum detector count rate is CR_{max}, the minimum scan time such that the detector counts all incident photons is equal to

$$T_{det} = \frac{N_{raw} \cdot \exp\left(-\mu \cdot d\right)}{CR_{max} \cdot f_{time}} \tag{5}$$

In this analysis we conservatively limited the photon count rate to 3 million counts per second for rays that pass through the center of the breast. The attenuation of 50% glandular / 50% adipose tissue at 80 kVp was estimated from published attenuation values of breast tissues.[14]

2.4 Sampling Considerations

Assuming unlimited source power and detector count rate, the scan time ($T_{sampling}$) that provides sufficient azimuthal sampling is

$$T_{sampling} = (\text{scan time per source location}) \cdot (\text{number of source locations per view}) \cdot (\text{number of view angles}) \tag{6}$$

A method for determining the number of views for sufficient azimuthal sampling for the inverse geometry was previously published, therefore we only briefly summarize the method here.[7] The method calculates the radon-space sampling of the in-plane IGCT geometry, where the in-plane geometry consists of all rays that connect each source position on one source row to all pixels on the directly opposite detector row. For each ray, the two radon-space parameters, ρ and θ are calculated, where ρ is the perpendicular distance of the ray to isocenter, and θ is the rotation angle of the ray. Radon space is defined by ρ spanning the field of view and θ from zero to 2π. Each ray samples one point in radon space, and the collection of rays connecting the IGCT source row to the detector row sample a slanted swath in radon space. When the gantry rotates, the ρ values remain unchanged while the θ values are shifted by the gantry rotation angle, thereby resulting in a new sampled swath. By investigating the radon space sampling, we can determine that minimum number of views that provide complete sampling.

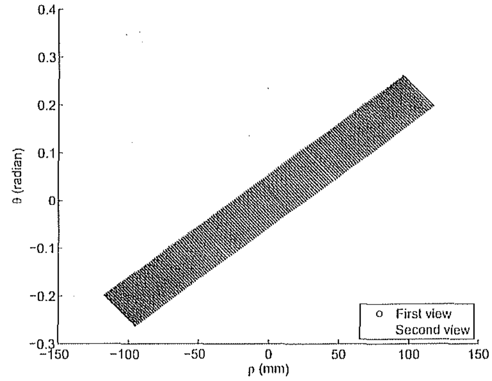

Figure 3. The radon-space sampling of two consecutive IGCT views when the angular distance between views is $2\pi/55$ radian.

Table 2. Parameters used to calculated the minimum scan times

Breast Diameter (cm)	mAs_{cone} (mAs)[13]	Photon Fluence F (photons/mm^2)[13]	N_{raw} (photons)
10	10.7	3.7×10^7	7.7×10^6
14	56.6	1.8×10^8	3.9×10^7
18	135.8	4.4×10^8	9.3×10^7

2.5 Minimum Scan Time

In the previous sections, we considered the constraints of source power, detector count rate, and sampling separately. The minimum scan time that meets all three constraints (T_{min}) is

$$T_{min} = max(T_{source}, T_{det}, T_{sampling}) \tag{7}$$

3. RESULTS

For the IGCT system specified in Table 1, the voxel at isocenter is irradiated for 5% of the total scan time, (Eq. 1) and the photon flux per mA is increased by a factor of 3.7 due to the transmission source and inverted geometry (Eq. 2). The number of incident raw photons per IGCT detector pixel (N_{raw} in Eq. 3) is listed in Table 2, along with the technique factors that provide dose equivalent to mammography for cone-beam breast CT systems. The radon space analysis determined that 55 was the minimum number of views that provided sufficient azimuthal sampling. Figure 3 shows the radon-space sampling of two consecutive IGCT views when the total number of views over 2π is 55. There are no gaps between the swaths corresponding to the two views, demonstrating that sufficient azimuthal sampling is achieved. Table 3 summarizes the minimum IGCT scan times that meet the source, detector, and sampling constraints for the three breast diameters. The fifth column lists the minimum scan time that meets all three constraints.

Table 3. Minimum scan times for each breast diameter

Breast Diameter (cm)	T_{source} (s)	T_{det} (s)	$T_{sampling}$ (s)	T_{min} (s)
10	0.2	6.5	0.8	6.5
14	1.1	14.3	0.8	14.3
18	2.5	14.7	0.8	14.7

4. DISCUSSION AND CONCLUSIONS

The minimum IGCT scan time ranged from 6.5 to 14.7 seconds depending on breast size. These scan times are comparable to those of proposed cone-beam breast CT systems (e.g., 17 - 33 seconds[15]). For all breast sizes, the IGCT scan time is limited by the detector count rate. For example, for the 14-cm-diameter breast, the minimum scan time that met the source power limitation was 1.1 seconds, and the minimum scan time that achieved sufficient sampling was 0.8 seconds. The scan time can be reduced by increasing the detector count rate and/or increasing the detector area (which increases the fraction of time each voxel is scanned). In addition, an effective bowtie filter will be required to prevent saturation at the edges of the object.

Overall, this work supports preliminary feasibility of an inverse-geometry dedicated breast CT system. The inverse geometry has the potential for improved dose efficiency, an important benefit for dedicated breast scanning.

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