Radiation doses in volume-of-interest breast computed tomography—A Monte Carlo simulation study

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Purpose: Cone beam breast computed tomography (breast CT) with true three-dimensional, nearly isotropic spatial resolution has been developed and investigated over the past decade to overcome the problem of lesions overlapping with breast anatomical structures on two-dimensional mammographic images. However, the ability of breast CT to detect small objects, such as tissue structure edges and small calcifications, is limited. To resolve this problem, the authors proposed and developed a volume-of-interest (VOI) breast CT technique to image a small VOI using a higher radiation dose to improve that region's visibility. In this study, the authors performed Monte Carlo simulations to estimate average breast dose and average glandular dose (AGD) for the VOI breast CT technique.

Methods: Electron–Gamma–Shower system code-based Monte Carlo codes were used to simulate breast CT. The Monte Carlo codes estimated were validated using physical measurements of air kerma ratios and point doses in phantoms with an ion chamber and optically stimulated luminescence dosimeters. The validated full cone x-ray source was then collimated to simulate half cone beam x-rays to image digital pendant-geometry, hemi-ellipsoidal, homogeneous breast phantoms and to estimate breast doses with full field scans. 13-cm in diameter, 10-cm long hemi-ellipsoidal homogeneous phantoms were used to simulate median breasts. Breast compositions of 25% and 50% volumetric glandular fractions (VGFs) were used to investigate the influence on breast dose. The simulated half cone beam x-rays were then collimated to a narrow x-ray beam with an area of 2.5×2.5 cm² field of view at the isocenter plane and to perform VOI field scans. The Monte Carlo results for the full field scans and the VOI field scans were then used to estimate the AGD for the VOI breast CT technique.

Results: The ratios of air kerma ratios and dose measurement results from the Monte Carlo simulation to those from the physical measurements were 0.97 ± 0.03 and 1.10 ± 0.13 , respectively, indicating that the accuracy of the Monte Carlo simulation was adequate. The normalized AGD with VOI field scans was substantially reduced by a factor of about 2 over the VOI region and by a factor of 18 over the entire breast for both 25% and 50% VGF simulated breasts compared with the normalized AGD with full field scans. The normalized AGD for the VOI breast CT technique can be kept the same as or lower than that for a full field scan with the exposure level for the VOI field scan increased by a factor of as much as 12.

Conclusions: The authors' Monte Carlo estimates of normalized AGDs for the VOI breast CT technique show that this technique can be used to markedly increase the dose to the breast and thus the visibility of the VOI region without increasing the dose to the breast. The results of this investigation should be helpful for those interested in using VOI breast CT technique to image small calcifications with dose concern. © 2015 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4921069>]

Key words: cone beam breast computed tomography, average glandular dose, Monte Carlo simulation, volume-of-interest

1. INTRODUCTION

Dedicated cone beam breast computed tomography (breast CT), which produces nearly isotropic, high-resolution three-dimensional (3D) image data sets of the breast, has been developed and investigated over the past decade and is a promising x-ray imaging technique for the screening, diagnosis, and staging of breast cancers. With breast CT, the patient lies prone on a table with one of her breasts protruding downward through an opening and scanned by a rotating x-ray source-detector gantry

underneath the table. With this technique, only the breast is directly exposed to the x-rays, minimizing the radiation dose to the rest of the body. With the entire breast scanned by the detector, a single rotation or a half rotation plus detector coverage around the breast results in complete projection data for image reconstruction. A number of studies have been conducted to investigate various technical parameters in breast CT, including the x-ray tube,¹ x-ray spectra,^{2,3} digital detectors,⁴ scatter reduction/correction,^{5–11} image reconstruction algorithms,^{12,13} radiation doses,^{14,15} and operating techniques.² Various breast

FIG. 1. (a) Schematic drawings of breast-centered scans of either the full field (long purple dashes), i.e., no VOI mask present, or the VOI field (short blue dashes), i.e., with the VOI mask present. A moving VOI mask is used to track and expose the VOI region during the VOI scans. (b) Schematic drawings of VOI-centered scans of either the full field (long purple dashes) or the VOI field (short blue dashes). The center of the VOI region is located at the isocenter, and a stationary VOI mask is used during the VOI scans.

CT imagers have been developed and evaluated in clinical applications.^{16–18} Preliminary clinical studies have shown that breast CT produces similar results to mammography in the detection of soft tissue masses with similar average glandular doses (AGDs) to the breast but better lesion conspicuity.¹⁷ However, the ability of breast CT to image small calcifications, which have been used as an indicator for early stage breast cancer, was not superior to that of mammography in these studies.

Although the linear attenuation coefficients of calcifications are higher than those of breast tissue structures, small calcifications may have lower contrast owing to partial volume effects. As a result, higher exposures may be required to reduce the noise level and allow small calcifications to be seen. However, it is generally recognized that the AGD for a full field breast CT scan (referred to as a full field scan hereafter) should be kept as low as possible while maintaining acceptable image quality in clinical applications. Therefore, the AGD could increase to an unacceptably high level if the exposure level is allowed to increase to allow for imaging small calcifications.

To improve the visibility of small calcifications without increasing the AGD, we have proposed and investigated the volume-of-interest (VOI) breast CT technique.^{19,20} A similar technique was previously proposed and investigated for neuro-radiological applications by Chityala *et al.*^{21,22} With their technique, a specially designed filter with an opening placed between the x-ray source and the object to allow full x-ray exposures to be delivered to the VOI region and exposures to regions outside the VOI attenuated by the filter. Chen *et al.* reported on an approach to VOI breast CT (Ref. 19) in which a collimator-type VOI mask was used to deliver higher exposures to the VOI region while a full field scan with lower exposure was used to provide projection data outside the VOI regions. Chityala *et al.*²¹ and Chen *et al.*²³ have reported as high as 10 times standard full field scan exposure can be applied to the VOI region, and the x-ray exposure outside the VOI can be as low as 1/10 of that of a standard full field scan, keeping an overall

dose to the entire breast within that of a standard full field scan. Using high exposures to acquire projection data through the VOI region has improved the contrast-to-noise ratios in the reconstructed images of the VOI region thus enhanced the visibility of low-contrast objects inside the VOI.²⁰

Two scanning geometries have been proposed and investigated for implementing VOI breast CT techniques with the collimation approach: breast-centered and VOI-centered scans, as shown in Fig. 1.^{24,25} With a breast-centered scan, the center of the breast is aligned with the rotating axis and the VOI mask is moved during the VOI field scan to track and direct the uncollimated/filtered x-ray beam to expose the VOI region (short blue dashes). With a VOI-centered scan, the breast is positioned to align the central axis of the VOI region with the rotating axis to allow the VOI mask and the uncollimated x-ray beam to remain stationary on the x-ray source-detector gantry during the VOI field scan. Projection data obtained from the full field scan (long purple dashes) with low exposure are combined with the data from the VOI field scan with higher exposure to form a complete set of projection data for image reconstruction using the Feldkamp filtered back-projection algorithm.¹²

In addition to higher exposure, higher spatial resolution for VOI field scans may help improve the detection and visualization of small calcifications. A small-field high-resolution detector can be used to acquire projection data through the VOI in a VOI field scan, and a full field scan can be performed with a separate non-high-resolution detector. Alternatively, if the high-resolution detector has a large enough field of view to cover the entire breast, then that detector may be used to perform both the high-resolution VOI field scans and the non-high-resolution full field scans.

VOI breast CT may be useful for diagnostic workup if a suspicious lesion is detected by screening mammography. In diagnostic workup, full field mammograms may be acquired again to provide scout images that can be used to look for suspicious regions prior to performing spot/magnification

mammography and/or other diagnostic imaging modalities. For this task, a full field breast CT scan may be used in replacement of full field mammography and a VOI field breast CT scan in replacement of spot/magnification mammography. The full field breast CT images may be used as scout images to determine the location of the suspicious lesions (i.e., selection of VOI) along with the screening mammograms based on their spatial relationship relative to the breast anatomy. Because it is not required to achieve the same high image quality throughout the entire breast for the scout images, either a standard or a low-exposure full field scan may be performed. A high-exposure VOI field scan may be performed to achieve high image quality in the suspicious region to provide a more clear and detailed look for more accurate characterization of the lesion. VOI breast CT may also be useful for surgical planning to evaluate tumor margin and for treatment planning to monitor the growth or recession of the tumor.

Despite the improved visualization and detection of small calcifications with VOI breast CT techniques, it is necessary to investigate the radiation dose from VOI breast CT compared with the dose from a standard full field scan. To this end, we performed Monte Carlo simulations to estimate the normalized average breast dose and normalized AGD for full field and VOI field scans of breast models that comprised homogeneous mixtures of glandular and adipose tissues. The results of the full field and VOI field scans were then used to estimate AGDs for the VOI breast CT technique, and those estimates were compared with doses from standard full field scans.

2. MATERIALS AND METHODS

2.A. Physical measurements

2.A.1. Bench-top breast CT scanner

Physical measurements to validate our Monte Carlo estimates were performed using a bench-top experimental breast CT scanner built in our laboratory (Fig. 2). This system consisted of a general radiographic x-ray unit, an amorphous silicon/cesium iodide flat panel detector (PaxScan 4030CB, Varian Medical Systems, Salt Lake City, UT), and a rotating table. The x-ray unit is composed of an x-ray generator (Indico 100 SP, Communications & Power Industries, Ontario, Canada) coupled to an x-ray tube (G1592BI, Varian Medical Systems) with nominal focal spot sizes of 0.6 and 1.2 mm.

FIG. 2. The bench-top experimental breast CT scanner was used to perform physical measurements.

The operating range of the x-ray unit was up to 150 kV for radiographic use and 125 kV for fluoroscopic use. The rotating table was positioned at the isocenter and was used to place and rotate phantoms driven by a stepping motor (B4872TS, Velmex, Inc., Bloomfield, NY) for a 360° scan. The distance between the x-ray source and the isocenter was 75 cm. The breast CT scanner was operated at 80 kVp on the basis of the results in our previous study.²⁶

2.A.2. Physical cylindrical phantoms

Three cylindrical polycarbonate phantoms with different diameters (11, 13, and 15 cm) but the same height (9 cm) were constructed for air kerma and dose measurements. Two “through-holes,” one at the phantom center and the other at 4.5 cm from the center (representing the VOI center), were opened in each phantom to accommodate an ion chamber and optically stimulated luminescence (OSL) dosimeters. The centers of the phantoms were placed on the rotating table and aligned with the focal spot plane.

2.A.3. Air kerma and dose measurements

The experimental setup is illustrated in Fig. 3(a). A full cone x-ray beam was used to perform physical measurements. The field-of-view of the x-ray beam was similar to the dimensions of the flat panel detector at the detector plane, resulting in a fan angle (θ_{fan}) of about $\pm 11.3^\circ$ and a cone angle (θ_{cone}) of $\pm 8.5^\circ$. Air kerma measurements were performed using an ion chamber (10X6-6, Radcal Corporation, Monrovia, CA) associated with an ion chamber digitizer (model 9660) connected to an Accu-Pro control unit (Radcal Corporation). The ion chamber was placed in the center through-hole, and a polycarbonate rod was placed in the other through-hole [Fig. 3(b)] to measure the air kerma rate (mR/s) at each phantom center. With the circular symmetry of the phantoms, there was no need to rotate the phantoms to perform a 360° scan while measuring exposure rates. The mA setting was varied with the exposure time fixed to determine the linear relationship between the measured exposure rates (mR/s) and mAs, whose slope was then computed to estimate mR/mAs. Air kerma rates were also measured at the same location with the phantoms removed. The measured air kerma rates were used to calculate air kerma ratios, defined as the ratio of the air kerma rate with the phantom to the air kerma rate free-in-air. The air kerma ratios were averaged over all mA settings for each phantom.

Point dose was measured in the 13-cm phantom with both the breast-centered and the VOI-centered scans. Two nano-Dot OSL dosimeters (Landauer, Inc., Glenwood, IL) were each sandwiched between two polycarbonate rods and then placed in the two through-holes [Fig. 3(c)]. The phantom was rotated 360° while being exposed to full cone x-rays using standard imaging technique for full field scans. The exposed OSL dosimeters were then read out with a specially designed reader (microStar, Landauer, Inc.), which used a light-emitting diode array to stimulate the dosimeter material and release the stored x-ray photon energy into light signals, which were converted into doses (in mGy) using the calibrated OSL

FIG. 3. (a) Schematic drawings of full cone x-rays for physical measurements. The center of each cylindrical phantom was positioned at the isocenter of the breast CT scanner and aligned with the focal spot plane. (b) To measure air kerma rates, an ion chamber was placed in the center hole of each cylindrical phantom and a polycarbonate rod was placed in the other through-hole. (c) To measure point doses, OSL dosimeters and polycarbonate rods were placed in both the through-holes of the 13-cm phantom.

dosimeters. The doses measured in the breast-centered and the VOI-centered scans were compared with the dose estimates from the simulations.

2.B. Monte Carlo simulation

We simulated breast CT with Monte Carlo methods, which track photons from a randomly generated distribution for their positions, movement, interactions, and energy deposited in the attenuating object. The Electron–Gamma–Shower system code (EGSnrc)²⁷ is a package developed by the National Research Council of Canada for Monte Carlo simulation of coupled electron–photon transport in an arbitrary geometry for particles with energies ranging from a few keV up to several hundred GeV. BEAMnrc and DOSXYZnrc are two user codes^{28,29} based on EGSnrc Monte Carlo simulation; BEAMnrc is used to simulate radiation sources, and DOSXYZnrc is designed for dose calculation in 3D voxel geometry. The output from BEAMnrc, referred to as “phase space file” in the code, is a summary of data about particles crossing a scoring plane, including particle number, positions, directions, and energies. The BEAMnrc output can be used as an x-ray source in DOSXYZnrc code for subsequent dose estimation.

2.B.1. Digital cylindrical phantoms for validation

Three digital phantoms with a voxel size of $1 \times 1 \times 1 \text{ mm}^3$ were created in 3D CT data format related to DOSXYZnrc/EGSnrc to mimic the physical cylindrical phantoms in Sec. 2.A.2. An enclosed air channel was created to replace the center through-hole at each phantom center. All phantoms were centered at the isocenter for performing breast-centered scans and estimating air kerma at the isocenter. An additional digital phantom with no air channel was created to simulate the 13-cm polycarbonate phantom for dose estimates. This digital

phantom was also centered at the isocenter to allow for breast-centered scans. To simulate VOI-centered scans, this digital phantom was “moved” 4.5 cm along x -direction in the CT data to center the VOI region at the isocenter.

2.B.2. Digital hemi-ellipsoidal phantoms

Digital hemi-ellipsoidal phantoms with a voxel size of $1 \times 1 \times 1 \text{ mm}^3$ were created to simulate pendant-geometry, uncompressed breasts [Fig. 4(a)]. Breast is mainly composed of glandular and adipose tissues. To study the effect of breast composition, the hemi-ellipsoidal phantoms were configured to simulate two different homogeneous breast tissue compositions to represent a typical range of volumetric glandular fractions (VGFs) in breasts, i.e., ratios of glandular tissue volume to the entire breast volume. Vedantham *et al.* reported that VGF ranged from 1.2% to 72% with a mean value of 17%.³⁰ Huang *et al.* reported that mean VGFs were about 18.5% for bra cup sizes A and B and 13% for sizes C and D,³¹ and Yaffe *et al.* reported that breast VGF was up to 55% with a mean value of 14.3% when the skin was excluded.³² These studies led us to select breast compositions of 25% and 50% VGFs for constructing the digital breast models. Breast composition is commonly expressed as the glandular tissue component with weight percentage; thus, for a breast with a VGF of V_g , the glandular fraction by weight (W_g) and the breast density (ρ_{breast}) were calculated by the following equations:

$$W_g = \left\{ \frac{(1 - V_g) \rho_a}{V_g \rho_g} + 1 \right\}^{-1}, \quad (1)$$

$$\rho_{\text{breast}} = \rho_g V_g + \rho_a (1 - V_g), \quad (2)$$

where ρ_g and ρ_a are the densities of glandular and adipose tissues, respectively, and V_g and $1 - V_g$ are the volume fractions of glandular and adipose tissues in the breast, respectively.

FIG. 4. (a) Schematic drawings of half cone beam x-rays for breast CT scans of either the full field (long purple dashes) or the VOI field (short blue dashes). (b) Dose gradient at the top slice resulting from a projectional full field exposure. (c) Dose gradient at the central slice resulting from a projectional VOI field exposure.

The densities of glandular and adipose tissues have been previously reported to be 1.04 and 0.93 g/cm³,³³ respectively. We estimated that breasts with 25% and 50% VGFs have breast densities of 0.958 and 0.985 g/cm³, respectively, and have glandular tissue components of 27% and 53% by weight, respectively. The attenuation coefficients of these two breast tissues for Monte Carlo simulation were computed using the elemental composition and the density of glandular tissue from Hammerstein *et al.*³³ and those of adipose tissue from the National Institute of Standards and Technology (NIST).³⁴

In addition to breast composition, breast size also affects the breast dose. The effective breast diameters from a breast CT images studied by Huang *et al.* were between 11 and 13.5 cm with effective breast lengths between 5.7 and 9.7 cm.³¹ Vedantham *et al.* study reported a mean effective breast diameter of 13.4 cm with mean effective breast length of 9.5 cm.³⁰ Boone *et al.* investigated the effect of the breast length in absorbed dose and reported no significantly different findings between breast lengths ranging from $D/2$ to D for breast diameter of D .¹⁴ Based on these studies, a standard uncompressed breast with a diameter of 13 cm and a length of 10 cm was employed for the breast models in this study.

2.B.3. Spectrum model

Our BEAMnrc Monte Carlo code was previously configured and validated to generate photons simulating 80 kVp x-rays.³⁵ X-ray filtration was adjusted to achieve a half-value layer (HVL) of 4.08 mm, matching the values measured for the x-rays used in our experimental study. A scoring plane was defined at the exit of the collimator to create and simulate a full cone x-ray source emitted from the focal spot, which was located at 40 cm from the isocenter.

2.B.4. Monte Carlo calculation

2.B.4.a. Validation using phantom studies. The imaging geometry in Fig. 3(a) was simulated in the DOSXYZnrc code to compare the physically measured air kerma ratios and point doses. Propagation and energy deposition of x-rays in the digital cylindrical phantoms with an enclosed air channel were simulated for each breast CT scan, which included 300 views evenly spread over a 360° revolution around the individual phantoms. For each digital phantom, a total of 10⁹ photons were tracked and the energies deposited in the phantom voxels were tallied. The air doses were then estimated both with and without the phantoms present. The estimated air doses were used to calculate air kerma ratios (the ratio of the air dose with the phantom present to the air dose without the phantom present) for individual phantoms. The estimated air kerma ratios were then compared with the measured air kerma ratios. Propagation and energy deposition in the digital phantoms with no enclosed air channel were also simulated. A total of 10⁹ photons were tracked for each scan and the tallied energy depositions were used for dose estimates for both breast-centered and VOI-centered scans. The estimated point doses for both scanning techniques were compared to each other and compared with the physical measurements.

2.B.4.b. Dose estimations for breast CT. The imaging geometry in Fig. 4(a) was simulated in the DOSXYZnrc code to estimate breast dose with full field and VOI field breast CT scans. A 3-mm thick lead plate was used to collimate the simulated full cone x-ray source into a half cone x-ray source to simulate full field breast CT scans (long purple dashes) where the cone angle ranged from -1.1° to 15.6° while the fan angle remained at ±11.3°. The homogeneous, hemi-ellipsoidal digital phantoms were centered on the rotating axis, and the top

of the digital phantom was aligned with the focal spot plane. To simulate VOI field coverage (short blue dashes), a 3-mm thick lead mask with an opening was incorporated into the code to collimate the simulated half cone x-ray source into a narrow x-ray beam with an area of $2.5 \times 2.5 \text{ cm}^2$ at the isocenter plane. The center of the VOI opening corresponded to the geometric center of the hemi-ellipsoidal digital phantom in this study.

Because of the circularly symmetric geometry of the homogeneous, hemi-ellipsoidal phantoms, exposure with a 360° breast CT scan results in a radially symmetric radiation dose gradient within the phantom for each given z -slice. That is, all of the points at a certain radius from the phantom center received the same radiation dose. On the basis of this property, a projectional radiographic exposure at one x-ray source position was simulated, and the resulting dose gradient was then used to simulate multiple projectional radiographic exposures from different source positions. The simulation results were used to calculate radiation doses from the same radial distances to obtain radial dose gradients for a 360° breast CT scan.

Between 10^6 and 10^7 photons were used to image the hemi-ellipsoidal phantoms, and the tallied energy depositions were used for dose gradient estimates for full field [Fig. 4(b)] and VOI field [Fig. 4(c)] radiographic exposures. The number of photons used for the projectional radiographic exposure was proportional to the 10^9 photons used for 300 views over a 360° scan. Four projectional radiographic exposures were performed for full field and VOI field, and individual dose results were calculated and averaged between all four. Then, the radial dose gradient $D_n(r)$ for individual z -slices was computed using Eq. (3)

$$D_n(r) = \sum_{i=1}^n D(r, \theta_i), \quad (3)$$

where $\theta_i = \theta_0 + (2\pi/n) \times (i-1)$, θ_0 is the initial sampling angle, and n is the sampling number of projection views. The θ_0 ranged from 0° to 80° with an increment of 10° and the n was between 4 and 300. Then, all the resulting slice-section radial dose gradients were stacked together to generate a two-dimensional (2D) radial dose gradient to determine the 3D absorbed dose distributions in a voxel-based breast phantom and to estimate average absorbed dose $\bar{D}_n$ using

$$\bar{D}_n = \frac{\sum_{i=1}^N D_{n,i} \times \rho_i \times V}{\sum_{i=1}^N \rho_i \times V}, \quad (4)$$

where D_i and ρ_i are the absorbed dose and tissue density, respectively, for the i th voxel; V is the volume of an individual voxel (1 mm^3 in this study); and N is the total number of voxels in the entire breast phantom. The average breast doses were then calculated for all of the possible combinations of different θ_0 and n values. The standard deviation of dose estimates, defined as the ratio of the standard deviation to the average value ($\sigma/\bar{D}$), was computed to investigate the effects of the θ_0 and the n values on breast dose variation.

2.B.5. Conversion of average breast doses to AGDs

The assessment of breast dose is performed as part of the quality assurance protocol for x-ray examinations, and AGD is commonly used as the standard dose index in clinical practice. The estimated average dose delivered to the breast tissue can be used to compute the average dose to the glandular tissue component of the breast using the energy-dependent ratio of the mass energy absorption coefficient of glandular tissue to that of the breast tissue, expressed by the following equation:

$$\bar{D}_g = \bar{D} \times (\mu_{\text{en}}/\rho)_{\text{breast}}^{\text{glandular}}, \quad (5)$$

where $\bar{D}$ and $\bar{D}_g$ are the average breast dose and AGD, respectively. In this study, $(\mu_{\text{en}}/\rho)_{\text{breast}}^{\text{glandular}}$ for 25% and 50% VGF breasts were determined to be 1.27 and 1.16, respectively, on the basis of the following: (1) the mean energy of the x-ray spectrum or spectrum model both for physical measurements and for Monte Carlo simulation was assumed to be 47 keV, (2) the elemental compositions of adipose and glandular on the basis of a report by Hammerstein *et al.*³³ were used to calculate the elemental compositions of breast tissues with 25% and 50% VGFs, and (3) the mass energy absorption coefficients of the elemental media based on the published data by the NIST (Ref. 34) were used to calculate the mass energy absorption coefficients of glandular tissue and breast tissues with 25% and 50% VGFs.

Equations (3)–(5) were applied to both full field and VOI field breast CT scans to calculate the average breast doses and the AGDs over the entire breast and over the VOI region. For dose analysis, we estimated the AGDs for simulated breast CT scans—both full field and VOI field—with different sampling numbers of projection views.

2.B.6. Quasi-homogeneous breast phantoms with full field breast CT scans

Because glandular tissue is not separated from adipose tissue in the voxels for a homogeneous breast model, AGD cannot be directly estimated from Monte Carlo simulation. Instead, dose deposited to each voxel is converted into glandular dose with the F -factor using Eq. (5) to estimate the AGD. To investigate the accuracy in AGD estimations obtained from Eq. (5) along with the assumptions for the homogeneous breasts, we constructed digital quasi-homogeneous hemi-ellipsoidal phantoms, where glandular tissue voxels (dark gray) and adipose tissue voxels (gray) were evenly distributed with specific patterns to simulate 25% and 50% VGF breasts in the Monte Carlo simulations (Fig. 5). Imaging geometry, simulated x-ray source, and breast dimensions in Fig. 4(a) were used to perform full field breast CT scans. A total of 10^9 photons were tracked to simulate 300 evenly spaced views over 360° . Energies deposited in the glandular tissue voxels were added up and then divided by total glandular tissue masses to calculate AGDs. The results were compared with those estimated for the homogeneous breast models using Eq. (5).

FIG. 5. Quasi-homogeneous breast models were used to estimate normalized AGD for (a) 25% and (b) 50% VGF hemi-ellipsoidal phantoms.

2.C. Dose estimation for the VOI breast CT technique

The AGD for the VOI breast CT technique ($AGD_{\text{voi-BCT}}$) was calculated as a simple addition of the AGDs for a full field scan and for a VOI field scan with weighting factors to account for the different exposure levels used for each scan as follows:

$$AGD_{\text{voi-BCT}} = \alpha \times AGD_{\text{ff}} + \beta \times AGD_{\text{voi}}, \quad (6)$$

where AGD_{ff} and AGD_{voi} are the acceptable AGDs for the full field and VOI field scans, respectively, using a standard exposure level technique; and α and β are the weighting factors representing the lower exposure level for full field scans and the higher exposure level for VOI field scans, respectively (with $\alpha \leq 1$ and $\beta \geq 1$). The AGD for the VOI breast CT technique was then compared with the AGD for a standard full field scan, and the ratio was expressed with the following equation:

$$\frac{AGD_{\text{voi-BCT}}}{AGD_{\text{ff}}} = \alpha + \beta \times \frac{AGD_{\text{voi}}}{AGD_{\text{ff}}}. \quad (7)$$

All measured and estimated doses were normalized to the free-in-air dose at the isocenter and presented as dose (mGy) per unit of air kerma (mGy) for comparison.

3. RESULTS

3.A. Simulation validation using physical phantoms

Table I lists the air kerma ratios from the Monte Carlo estimates and the physical measurements. Both showed that

TABLE I. Air kerma ratios from the Monte Carlo simulations and the physical measurements.

Measurement type	Phantom size		
	11 cm	13 cm	15 cm
Simulation	0.662	0.599	0.570
Physical measurement	0.734	0.619	0.555
Simulation/physical measurement	0.902 (0.022)	0.968 (0.025)	1.026 (0.027)

Note: The numbers in parentheses are standard deviations.

the air kerma ratio decreased with increasing phantom size because of more x-ray attenuation at the phantom center in larger diameter phantoms. The ratios of the Monte Carlo estimates to the measured values were between 0.90 and 1.03 with an average of 0.97 ± 0.03 , indicating that the accuracy of the Monte Carlo simulation was adequate.

Table II shows the normalized point doses at four radial positions (0, 1.5, 3, and 4.5 cm from the phantom center) from the Monte Carlo estimates and at two positions (0 and 4.5 cm from the center) from the physical measurements for both breast-centered and VOI-centered scans. The Monte Carlo estimates of dose level increased with the distance of the radial position from the phantom center for both the breast-centered and the VOI-centered scans, and the physical measurements showed that the position closer to the phantom edge had higher doses than the position at the phantom center for both types of scans. The ratio of the VOI-centered doses to the breast-centered doses at the corresponding radial positions ranged from 0.98 ± 0.10 to 1.00 ± 0.11 for the Monte Carlo estimates and was 1.02 ± 0.04 at both positions for the physical measurements, indicating that the VOI-centered and the breast-centered scans resulted in similar dose levels. The normalized point doses for the breast-centered and VOI-centered scans were averaged and used to calculate the ratio of the Monte Carlo estimates to the measured values. The calculated ratios were 1.10 ± 0.13 at the phantom center and 1.10 ± 0.12 at 4.5 cm from the center, confirming that the simulation methods were valid.

3.B. Dose variation vs sampling number of projection views and initial sampling angle

Figure 6(a) shows the normalized 2D radial dose gradients for the 25% VGF simulated breasts with a simulated 360° full field scan and a simulated 360° VOI field scan. The sampling numbers of projection views (n) were 4 and 120, and the initial sampling angle (θ_0) was 0° . The variation in breast dose distribution over the 2D radial dose gradients was not visually significant change with full field scans but was significant change with VOI field scans, especially close to the VOI boundaries, owing to the collimation from the VOI mask. Similar variations were observed at different θ_0 and n values

TABLE II. Normalized point doses (mGy/mGy) at different radial positions for the Monte Carlo estimates and the physical measurements.

Measurement type	Scan type	Distance from the phantom center			
		0 cm	1.5 cm	3 cm	4.5 cm
Monte Carlo estimate	Breast-centered	0.475	0.485	0.522	0.576
	VOI-centered	0.476	0.485	0.515	0.566
	VOI-centered/breast-centered	1.00 (0.11)	1.00 (0.11)	0.99 (0.11)	0.98 (0.10)
Physical measurement	Breast-centered	0.429			0.515
	VOI-centered	0.437			0.526
	VOI-centered/breast-centered	1.02 (0.04)			1.02 (0.04)
Monte Carlo estimate/physical measurement		1.10 (0.13)			1.10(0.12)

Note: The ratios of the VOI-centered scans to the breast-centered scans were calculated for the Monte Carlo estimates and the physical measurements. The ratios of the Monte Carlo estimate to the physical measurements were computed using the average of the breast-centered and the VOI-centered values. The numbers in the parentheses are standard deviations.

as well as for the 50% VGF simulated breasts. The standard deviations of dose estimates were calculated to study the influence of θ_0 and n on the breast dose. With full field scans, the standard deviation was less than 1% for both 25% and 50% VGF simulated breasts, indicating that the θ_0 and n have minor influence on dose estimation. Figure 6(b) shows that with the VOI field scans, the standard deviation averaged from all θ_0 values decreased as n increased. This result indicates that dose variations caused by the θ_0 can be minimized (<2% or less) if 30 or more projection views are used for dose estimation. Figure 6(c) shows that with the VOI field scans, the standard deviation averaged from all n values was between 10% and 26% for different θ_0 values, and no correlation was observed between the standard deviation and θ_0 .

3.C. Normalized average dose vs sampling number of projection views

Figure 7 shows the normalized average dose plotted as a function of sampling number of projection views for the 25% and 50% VGF simulated breasts imaged with a full field scan and a VOI field scan. With full field scans, no substantial difference in normalized average dose was observed as n varied, but with VOI field scans, substantial dose differences were observed between n values of less than 30 and n values of at least 30. This implies that relatively few projection views, e.g., 30 views, can be used for dose estimation to decrease the simulation time with no substantial difference in breast dose.

3.D. 2D normalized breast dose profiles

Figure 8(a) shows the surface plots of the normalized 2D radial dose gradients to illustrate the dose distributions for both 25% and 50% VGF simulated breasts and both types of field scans when n is 300. Figures 8(b) and 8(c) were plotted as a function of radial position at the central slice ($z = 50$) and as a function of axial position at the rotating axis ($r = 0$), respectively, for both 25% and 50% VGF simulated breasts and both types of field scans. Doses from VOI field scans were substantially reduced at all locations, inside and outside the VOI region, compared with doses from full field scans.

3.E. Normalized average dose and normalized AGD

Table III lists the estimated normalized average dose for the 25% and 50% VGF simulated breasts imaged with a simulated full field scan and a simulated VOI field scan when n is 300 and θ_0 is 0° . With the simulated full field scans, the normalized average doses were estimated to be 0.60 and 0.63 mGy/mGy over the entire breast for the 25% and 50% VGF simulated breasts, respectively; and 0.57 and 0.59 mGy/mGy over the VOI region, respectively. With the simulated VOI field scans, the normalized average dose over the entire breast was substantially reduced to about $5.57\% \pm 0.07\%$ of the normalized average dose with the full field scans for both 25% and 50% VGF simulated breasts, and the normalized average dose over the VOI region was lowered to about half of the normalized average dose with the full field scans. In addition, normalized average dose increased with VGF percentage.

Table IV lists the estimated normalized AGD for the 25% and 50% VGF simulated breasts imaged with both types of simulated scans. With full field scans, the normalized AGDs were estimated to be 0.76 and 0.73 mGy/mGy over the entire breast for the 25% and 50% VGF simulated breasts, respectively, and were 0.72 and 0.69 mGy/mGy over the VOI region, respectively. Similar to the normalized average dose, the normalized AGD over the entire breast with VOI field scans was substantially reduced to $5.58\% \pm 0.08\%$ of that with full field scans, and the normalized AGD over the VOI region with VOI field scans was about half of that with full field scans.

The estimated normalized AGDs were 0.761 and 0.733 mGy/mGy over the entire breast for the 25% and 50% VGF quasi-homogeneous simulated breasts, respectively. The results are similar to those for the homogeneous simulated breasts using Eq. (5), indicating that correction with the F -factor along may be used to estimate AGD for homogeneous breast models with adequate accuracy.

3.F. Normalized AGD for the VOI breast CT technique

Table V lists the AGDs for the VOI breast CT technique normalized to that for a standard full field scan for the 25%

FIG. 6. (a) Normalized 2D radial dose gradients, obtained from stacking slice-section radial dose gradients, for 25% VGF simulated breasts imaged with a simulated 360° full field scan and a simulated 360° VOI field scan at an initial sampling angle of 0°. The units for the scale bars are mGy/mGy. (b) The influence of the initial sampling angle (θ_0) on dose estimates with the VOI field scans was expressed as standard deviation and plotted as a function of the sampling number of projection views (n). (c) The influence of the sampling number of projection views (n) on dose estimates with the VOI field scans was expressed as standard deviation and plotted as a function of the initial sampling angle (θ_0).

and 50% VGF simulated breasts. Values of α varying from 0.05 to 1.0 and β varying from 1 to 16 were selected to calculate $AGD_{\text{VOI-BCT}}/AGD_{\text{FF}}$ in Eq. (7). The result showed that the normalized AGDs for the VOI breast CT technique can be kept at the same level as or lower than those for a standard full field scan with the exposure level for the VOI field scan increased by a factor of as much as 12.

4. DISCUSSION

In the air kerma ratio study, the Monte Carlo estimates were slightly lower than the physical measurements for the 11-cm phantom but similar to the measurements for the 13-cm and 15-cm phantoms, and in the point dose study, the Monte Carlo estimates were slightly higher than the physical measure-

FIG. 7. Normalized average breast dose plotted as a function of sampling numbers of projection views for the 25% and 50% VGF simulated breasts.

ments at both positions inside the 13-cm cylindrical phantom. There are several possible causes for these discrepancies. First, the modeled x-ray spectra for the Monte Carlo simulation had the same HVL as the x-ray spectra used in the experimental measurements. However, the same HVL does not imply that the two x-ray spectra were identical. Second, the OSL dosimeters placed in the cylindrical phantom were rotated with the phantom during the x-ray exposure. Further investigation is needed to find out whether the x-ray absorption in the dosimeters was affected by the rotating angle during the exposure. However, this is beyond of the scope of this study. Within the limits of the overall errors for both the Monte Carlo calculation and the physical measurements, our simulation model for the Monte Carlo method was a valid representation of the physical phantom.

13-cm, 10-cm long hemi-ellipsoidal digital phantoms were used to simulate median breasts for the Monte Carlo simulations of breast CT and to estimate normalized average doses and normalized AGDs with a full field scan and a VOI field scan. With full field scans, this study had results similar to those of our previous investigation,³⁵ which employed structured models (0.60 vs 0.59 and 0.63 vs 0.61 in normalized average dose for 25% and 50% VGF simulated breasts, respectively; 0.76 vs 0.74 and 0.73 vs 0.67 in normalized AGD for 25% and 50% VGF simulated breasts, respectively). This was expected because the previous investigation showed that hemi-ellipsoidal homogeneous models yield dose estimations similar to those of structured breast models if the homogeneous and structured models have the same VGF. The differences in normalized AGD estimates between structured models and homogeneous hemi-ellipsoidal phantoms were larger than the differences in normalized average dose, possibly due to the weight of spatial distribution of glandular tissue toward the central axis in real breasts. Because normalized AGDs are generally reported as a function of glandular fraction in breast, the influence of the spatial distribution of glandular tissues on breast dose has not been investigated. The results showed that the homogeneous and quasi-homogeneous breast models had higher normalized AGDs than the structured breast models. This is because breasts are mostly made up of

adipose tissues surrounding the tree-like structures consisting of a complex network of lobules and milk ducts (i.e., glandular tissues), and these glandular tissue structures are scatter-distributed within the breast but toward breast center. It is well known that breast dose increases with the radial distance from the breast center (i.e., breast dose at the breast center is the lowest). As a result, the total energy deposited in glandular tissue voxels from the structures breast model is lower than that in quasi-homogeneous breast model, resulting in lower normalized AGDs. Overall, this good agreement in dose estimations for standard full field scans provides additional validation for the Monte Carlo dose estimation between homogeneous and quasi-homogeneous models as well as between structured models and homogeneous hemi-ellipsoidal phantoms.

In this study, we performed Monte Carlo simulations using the EGSnrc package to estimate normalized AGDs for breast CT with full field and VOI field scans. A number of studies have been conducted to investigate normalized AGD for breast CT. Boone *et al.* used the SIERRA Monte Carlo code and employed cylindrical digital phantoms to compute normalized AGD as a function of x-ray spectrum, the use of bow-tie filter, breast size, and breast composition.¹⁴ Thacker and Glick used the GEANT3 Monte Carlo code and monoenergetic x-rays ranging from 10 to 120 keV to estimate normalized AGD for cylindrical breast phantoms with various combinations of breast size and breast composition.¹⁵ These studies and our study used similar simulation methods but different Monte Carlo software packages. Thus, it is possible to directly compare previous studies with ours to validate our results for full field scans. Here, we compared our results with those of Boone *et al.* on the basis of the same kVp (80 kVp) with no bow-tie filter. Using linear interpolation, Boone *et al.* reported that the normalized AGDs for 13-cm breasts with 25% and 50% VGFs were about 0.808 (vs 0.756) and 0.777 (vs 0.732) mGy/mGy, respectively. This study observed 6.5% and 5.8% lower AGDs than the AGDs reported by Boone *et al.* for 25% and 50% VGF breasts, respectively. These differences may be mainly caused by the differences in breast phantom models and x-ray beam quality between our simulation and theirs (model: hemi-ellipsoidal vs cylindrical and HVL: 4.08 vs 5.2 mm).

This study demonstrated VOI field scans have an advantage over full field scans. The normalized average doses and normalized AGDs for VOI field scans were substantially reduced by a factor of 2 over the VOI region and a factor of 18 over the entire breast. This result was obtained using the same exposure level (i.e., the same number of x-ray photons) for the full field scans and for the VOI field scans. This significant dose reduction results from the VOI mask, which attenuated x-ray exposure in the region outside the VOI region and reduced the scatter component both inside and outside the VOI region while the primary x-rays inside the VOI remained unchanged. We previously conducted an EGSnrc-based Monte Carlo simulation to estimate doses at two positions, one at the phantom center and the other at 4.5 cm from the phantom center (i.e., at the VOI center), and found that point doses were reduced by a factor of 1.6 at the VOI center and by a factor of about 7.5 at the phantom center (i.e., outside the VOI) for both breast-centered and VOI-centered scans.²⁵ There was a slight

FIG. 8. (a) Surface plots of normalized 2D radial dose gradients illustrate the dose distributions for the 25% and 50% VGF simulated breasts imaged with a simulated full field scan and a simulated VOI field scan with 300 projection views. (b) Normalized dose plotted as a function of radial position at the central slice (tan solid) for the 25% and 50% VGF simulated breasts with a full field scan and a VOI scan. (c) Normalized dose plotted as a function of axial position at the rotating axis (tan solid) for 25% and 50% VGF simulated breasts with a full field scan and a VOI scan.

difference in dose reduction over the VOI region between this previous study and the present study because different VGF percentages were used. Other reasons for the observed difference can be because the dose reduction was calculated based on a region close to the VOI center ($\sim 0.5^{\phi} \times 0.5^H$ cm) in the previous study but was calculated over the VOI region ($\sim 2.5^{\phi} \times 2.5^H$ cm) in this study. It is not surprising that the dose reduction over the VOI region is slightly different from that at the VOI center because the spatial dose varied. These two studies also both showed that breast compositions may

not significantly influence dose reduction factor over the VOI region.

Equation (7) provided the resulting AGDs for the VOI breast CT technique normalized to the AGD for the standard full field scans. Because researchers commonly keep the breast dose for each breast CT examination scan within the AGD limit for a full field scan, the value of $AGD_{\text{voi-BCT}}/AGD_{\text{ff}}$ was assumed to be no greater than one. We have reported that the dose reduction, the reciprocal of $AGD_{\text{voi}}/AGD_{\text{ff}}$ in Eq. (7), for the VOI field scan was by a factor of about 18 using a

TABLE III. Normalized average doses (mGy/mGy) for 25% and 50% VGF simulated breasts imaged with a simulated full field scan and a simulated VOI field scan on the basis of 300 projection views.

Glandular fraction	Region	Full field scan	VOI field scan
25% VGF	VOI region	0.567 (0.012)	0.297 (0.013)
	Entire phantom	0.595 (0.011)	0.045 (0.005)
50% VGF	VOI region	0.590 (0.013)	0.307 (0.014)
	Entire phantom	0.631 (0.011)	0.047 (0.005)

Note: The VOI region was at the geometric center of the simulated breast phantoms. The numbers in parentheses are standard deviations.

$2.5 \times 2.5 \text{ cm}^2$ VOI opening at the isocenter plane. On the basis of Eq. (7), such a significant dose reduction may lead to a β close to 18 whereas α can be as low as possible. This result indicates that to perform a complete VOI breast CT scan, the exposure level applied to VOI field scans can be about 18 times the exposure level for a standard full field scan to achieve higher quality images over the VOI region whereas the exposure level applied to full field scans can be as low as possible, close to the x-ray tube limit, to obtain a complete set of projection data for reconstruction. However, to stay within the AGD limit for each VOI breast CT examination from both scans, β can be up to 12 times the exposure level for performing a VOI field scan (Table V). We previously conducted a phantom study to investigate the VOI breast CT technique.²⁵ Two exposure levels were applied to performing VOI field scans—4 and 8 times the exposure level (i.e., β)—and 0.5 times the exposure level (i.e., α) was used for a full field scan. The image quality inside the VOI region was improved for both β values compared with the images acquired with a standard full field. Because of x-ray tube limit and heat loading concern, the mAs setting for 12 times the exposure level was not used to perform VOI field scan.

Normalized average dose and normalized AGD for a standard full field scan are dependent on breasts dimension and composition, x-ray spectral properties (such as target-filter combination, kVp, additional filtration, and HVL), and scanner geometry [such as source-to-isocenter distance (SID)]. Furthermore, the dose level for VOI field scans is dependent

TABLE IV. Normalized AGD (mGy/mGy) for 25% and 50% VGF simulated breasts imaged with a simulated full field scan and a simulated VOI field scan on the basis of 300 projection views.

Glandular fraction	Region	Full field scan	VOI field scan
25% VGF	VOI region	0.720 (0.016)	0.377 (0.017)
	Entire phantom	0.756 (0.014)	0.057 (0.006)
50% VGF	VOI region	0.685 (0.015)	0.356 (0.016)
	Entire phantom	0.732 (0.013)	0.055 (0.006)

Note: The VOI region was at the geometric center of the simulated breast phantoms. The numbers in parentheses are standard deviations.

on the dimensions of the VOI opening at the isocenter. The influence of the SID on normalized AGD is not significant because the estimated doses were normalized to the free-in-air dose at the isocenter. Although the AGD for full field scans have been investigated and reported by several studies, the present simulation study focuses on the AGD limit for the VOI breast CT technique. We have demonstrated that the AGD for the VOI breast CT technique is similar to or lower than that for a standard full field scan, thus minimizing the influence of breast specific factors, x-ray spectra, and VOI opening size on the normalized AGD for the VOI breast CT techniques. This study focuses on investigating doses for the VOI breast CT technique; however, the overall evaluation of the VOI breast CT techniques needs to include the image quality of VOI breast CT images.

In this study, we used the Monte Carlo simulation results for a single projectional exposure to simulate multiple projectional exposures from different source positions for a 360° rotation breast CT scan, to determine 3D dose distributions within the breast, and to calculate average breast doses. Results in Figs. 5 and 6 showed that the sampling number of projection views and the initial sampling angle significantly affected dose estimations for VOI field scans but had little effect for full field scans. These factors influenced dose in VOI field scans because the VOI mask limited the field of view [as shown in Fig. 4(a)], resulting in significant radial dose gradient variations in individual slice-sections, especially the slice-sections close to the VOI boundaries [Figs. 6(c) and 6(d)].

TABLE V. AGD (mGy) for the VOI breast CT technique normalized to the AGD (mGy) with standard full field scans for 25% or 50% VGF simulated breasts.

α value for low-exposure full field scan	β value for high-exposure VOI field scan								
	1	2	4	6	8	10	12	14	16
0.05	0.13	0.20	0.35	0.50	0.65	0.80	0.95	1.10	1.25
0.1	0.18	0.25	0.40	0.55	0.70	0.85	1.00	1.15	1.30
0.2	0.28	0.35	0.50	0.65	0.80	0.95	1.10	1.25	1.40
0.4	0.48	0.55	0.70	0.85	1.00	1.15	1.30	1.45	1.60
0.6	0.68	0.75	0.90	1.05	1.20	1.35	1.50	1.65	1.80
0.8	0.88	0.95	1.10	1.25	1.40	1.55	1.70	1.85	2.00
1.0	1.08	1.15	1.30	1.45	1.60	1.75	1.90	2.05	2.20

Therefore, when using fewer projection views for dose estimation with VOI field scans, dose estimates may vary with the initial sampling angle, and sampling more projection views reduces the dose estimation variations and the effect of the initial sampling angle becomes less significant. We have shown in Figs. 5(b) and 6(a) that the variations can be reduced to 2% with dose estimations kept at a similar level when 30 or more projection views were sampled. However, there are several limitations to the simulated breast approach. First, simulated breast phantoms have to be evenly symmetric. Second, homogeneous breast tissue is used for the symmetric phantom. Third, the VOI center is assumed to be on the rotating axis. Finally, simulated multiple projectional exposures obtained from the same projectional exposure may result in a correlated 3D dose distribution for a 360° rotation breast CT scan.

5. CONCLUSION

We conducted a Monte Carlo simulation study to estimate normalized average doses and normalized AGDs for full field and VOI field scans, and the estimated normalized AGDs from both types of scans were used to derive the normalized AGDs for the VOI breast CT technique. With the VOI breast CT technique, the doses to the entire breast can be kept at the same as or lower than those with a standard full field scan while the exposure level for the VOI field scan can be increased by a factor of as much as 12. This technique could be used to image suspicious regions in greater detail without increasing the dose to surrounding tissue to unacceptable levels and should be tested further to assess the image quality that can be obtained.

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