The effect of angular dose distribution on the detection of microcalcifications in digital breast tomosynthesis

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(Received 17 February 2010; revised 1 March 2011; accepted for publication 7 March 2011; published online 29 April 2011)

Purpose: Substantial effort has been devoted to the clinical development of digital breast tomosynthesis (DBT). DBT is a three-dimensional (3D) x-ray imaging modality that reconstructs a number of thin image slices parallel to a stationary detector plane. Preliminary clinical studies have shown that the removal of overlapping breast tissue reduces image clutter and increases detectability of large, low contrast lesions. However, some studies, as well as anecdotal evidence, suggested decreased conspicuity of small, high contrast objects such as microcalcifications. Several investigators have proposed alternative imaging methods for improving microcalcification detection by delivering half of the total dose to the central view in addition to a separate DBT scan. Preliminary observer studies found possible improvement by either viewing the central projection alone or combining all views with a reconstruction algorithm.

Methods: In this paper, we developed a generalized imaging theory based on a cascaded linear-system model for DBT to calculate the effect of variable angular dose distribution on the 3D modulation transfer function (MTF) and noise power spectrum (NPS). Using the ideal observer signal-to-noise ratio (SNR), d' , as a figure-of-merit (FOM) for a signal embedded in a uniform background, we compared the detectability of objects with different sizes under different imaging conditions (e.g., angular dose distribution and reconstruction filters). Experimental investigation was conducted for three different angular dose schemes (ADS) using a Siemens Novation^{TOMO} prototype unit.

Results: Our results show excellent agreement between modeled and experimental measurements of 3D NPS with different angular dose distribution. The ideal observer detectability index for the detection of Gaussian objects with different angular dose distributions depends strongly on the applied reconstruction filter as well as the imaging task. For detection tasks of small calcifications with reconstruction filters used typically in a clinical setting, variable angular dose distribution with more dose delivered to the central views may lead to higher d' than a uniform angular dose distribution.

Conclusions: The conspicuity of the detection of small calcifications may be improved, under certain imaging conditions, by delivering higher dose toward the central views of a tomosynthesis scan, while also reducing the dose at peripheral angles to keep total administered radiation dose equivalent. The degree of improvement depends on the choice of reconstruction filters as well as the imaging task. The improvement is more substantial for high-frequency imaging tasks and when an aggressive slice-thickness (ST) filter is applied to reduced the high-frequency noise at peripheral angles. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3570580]

Key words: tomosynthesis, breast imaging, microcalcifications, linear-system model, ideal observer, MTF, NPS, detectability

I. INTRODUCTION

Projection mammography is the current standard of breast cancer screening. However, it only provides a sensitivity and specificity of 89% and 85%, respectively.¹ This is in part due to the masking effect of overlapping breast tissue associated with the projection of three-dimensional (3D) anatomical data onto two-dimensional (2D) images. Recently, digital breast tomosynthesis (DBT) has been developed to overcome this limitation. In DBT, a number of projection images are acquired over a limited angular span and are reconstructed into a volume composed of thin slices (nominally

1 mm) oriented parallel to the detector plane, allowing the visualization of the breast at a specific depth while decreasing the obscuring effect of overlapping tissue. Various phantom and clinical studies have demonstrated the ability of DBT to increase breast lesion conspicuity.²⁻⁵

Minimizing the effect of overlapping tissue aids in the detection of masses, which is limited largely by breast tissue structural noise.⁶ The detection of microcalcifications, however, is to a greater degree limited by quantum and detector noise.⁶ Nishikawa *et al.*⁷ suggested that the reconstruction process in DBT effectively increases the electronic noise of

the final image due to increased projection numbers, possibly inhibiting microcalcification detection.

The studies by Nishikawa *et al.* and others have proposed a new method of DBT acquisition whereby two image acquisitions were performed: (1) a tomosynthesis scan with half the dose of screening mammography and (2) a single projection image made at the 0° (central) location with half the dose of screening mammography.^{7–9} Using this scheme, all views were used to reconstruct one DBT image volume for mass detection and the central, half-dose projection was analyzed by a computer-aided detection (CADe) scheme for microcalcifications detection. Das *et al.* expanded upon this technique and performed an observer localization receiver operating characteristic (LROC) study comparing the detection of masses and calcifications between the variable dose DBT (VD-DBT), with half the dose distributed to the central view, and the uniform dose DBT (UD-DBT) technique.^{8,9} Comparisons of reconstructed UD-DBT images with the central projection of the VD-DBT technique showed inferior microcalcification detection for VD-DBT. The study also found an insignificant difference for mass detection performance between the reconstructed images of the two techniques.

In the present work, a generalized physical image quality investigation was performed on the effect of angular dose distribution on lesion detectability in DBT. The aim of the study is to provide guidance to a tomosynthesis scan with variable dose distribution, where all projection views will be used to reconstruct a set of image slices for the detection of both masses and microcalcifications. We used a cascaded linear-system modeling approach to analyze the signal and noise characteristics of DBT for both uniform and variable angular dose distributions. The ideal observer signal-to-noise ratio (SNR) for the detection of Gaussian objects of different sizes was used as a figure-of-merit (FOM) to quantify the detectability of microcalcifications for both in-plane and 3D images. For experimental validation, two variable angular dose schemes (ADS) were implemented on a prototype DBT system to verify theoretical findings. In addition to quantitative measurement of signal and noise, the tissue-equivalent mammography phantom (model 011A, CIRS Inc., Norfolk, VA) was imaged to qualitatively verify our findings.

II. METHODS

II.A. Description of prototype DBT system

Our theoretical and experimental methods were developed based on a prototype Siemens Novation^{TOMO} DBT unit with a large area amorphous selenium (a-Se) full field digital mammography detector. (Caution: This is an investigation device and limited by US Federal law to investigational use. The information about this product is preliminary. The product is under development and is not commercially available in the US; its future availability cannot be ensured). The detector has 2816×3584 pixels with $85 \mu\text{m}$ pitch. The target/filter combination of Tungsten (W)/Rhodium (Rh) was used exclusively for tomosynthesis acquisition. During a typical DBT

FIG. 1. Diagram showing the geometry of the prototype DBT system and the angular range of different dose distributions (a) and (b) their corresponding contribution to the noise power spectrum in the frequency domain. In this example, a higher dose per view is delivered in the central region of the sampled space I_{cent} (with angular range θ_{IN}) than the peripheral regions I_{out-1} and I_{out-2} . The total angular range is designated as θ_{TOT} . The Hanning window width of the slice-thickness filter is represented by dotted lines (bottom).

scan, as shown in Fig. 1, the x-ray tube travels in a $\pm 25^\circ$ arc about a center of rotation (COR) located 4.5 cm above the detector cover, which is 1.5 cm above the a-Se layer. The source to imager distance (SID) is 65 cm and the source to COR distance (SAD) is 59 cm. The nominal angular span over which the x-ray exposures were made was approximately $\pm 22^\circ$ with 25 projection views using full detector resolution (i.e. “ $\times 25$ ” mode). In this mode, the detector performance is not degraded substantially by electronic noise while providing a reasonable scan-time of approximately 20 s for clinical imaging.¹⁰ It also provides maximum spatial resolution for the detection of microcalcifications. The angular position of the x-ray tube at the beginning of each x-ray exposure was measured with an inclinometer (MicroStrain, Inc., Burlington, VT) installed in

the tube housing and recorded in a parameter file. This file was later used in the reconstruction for accurate definition of imaging geometry.

For the present study, we have designated the coordinates in reconstructed 3D space as x , y , and z , as shown in Fig. 1(a), where x corresponds to the direction of tube travel, y is the direction perpendicular to x on the detector plane, and z is the slice-thickness direction. The central slice theorem dictates that the Fourier transform of the projection along angle θ_i corresponds to the frequency domain information along the same angle. We therefore introduce the polar coordinates, f_r and θ_i , which denote the radial frequency and the projection angle, respectively. The polar coordinates only apply to the axial cross-section (x - z) of the reconstructed space and may be converted to Cartesian coordinates through

$$\begin{aligned} f_z &= f_r \sin(\theta_i), \\ f_x &= f_r \cos(\theta_i), \\ f_r &= \sqrt{f_x^2 + f_z^2}. \end{aligned} \quad (1)$$

The prototype system has a partial-isocentric geometry (with a stationary detector); hence, the spatial frequencies of the projection images at angle θ_i , $f_{x'}$, and $f_{y'}$, are related to the reconstructed frequency domain through:

$$f_r = \frac{f_{x'}}{\cos(\theta_i)}, \quad f_y = f_{y'}. \quad (2)$$

Substituting Eq. (2) into Eq. (1), we obtain $f_{x'} = f_x$. The limited angular range of a tomosynthesis acquisition results in an incomplete sampling of the x - z frequency space, as seen in Fig. 1(b). Consequently tomosynthesis slices are usually reconstructed in the x - y plane, i.e., parallel to the stationary detector.

Tomosynthesis image reconstruction was performed using a modified filtered backprojection (FBP) algorithm, which was described in detail previously.^{11,12} Briefly, three reconstruction filters were used (Table 1): (1) Ramp filter (RA) with transfer function defined as

$$H_{RA}(f_r) = 2 \times \tan(\theta_{TOT}) \times \frac{|f_r|}{f_{r-NY}}, \quad \text{for } |f_r| \leq f_{r-NY}$$

with

$$f_r - NY = f_{NY} / \cos(\theta_i), \quad (3)$$

where θ_{TOT} is the angular range of DBT acquisition, and f_{NY} is the Nyquist frequency of the projection images (5.88 cycles/mm with full resolution readout); (2) Spectral apodization filter (SA) in the form of a Hanning window

$$H_{SA}(f_x) = 0.5 \left[1 + \cos \left(\frac{\pi f_x}{A} \right) \right], \quad (4)$$

where A defines the window width; and (3) slice-thickness filter (ST) in the form of a Hanning window

$$H_{ST}(f_z) = 0.5 \left[1 + \cos \left(\frac{\pi f_z}{B} \right) \right], \quad (5)$$

where B is the window width. Both A and B are given in multiples of f_{NY} .

II.B. Physical imaging performance of DBT using variable angular dose distribution

II.B.1. Theory: Cascaded linear-system analysis

The 3D cascaded linear-system model we developed previously for DBT^{13,14} was used to perform theoretical investigation of the presampling signal spectrum and NPS as a function of angular dose distribution.

Our analysis was generalized for DBT without focal spot motion. Hence, the same projection domain signal spectrum, $\Phi_p(f_{x'}, f_{y'})$, was used for all angular dose distributions. The signal and NPS of each projection view, $S_p(f_{x'}, f_{y'})$, were generated using a cascaded linear-system model for amorphous selenium (a-Se) flat panel detectors with model parameters chosen to match the measured detector performance of our prototype DBT system.¹⁵ The effect of scatter and its dependence on view-angle was not included in our present implementation of the model. The effect of this assumption will be discussed in Sec. III C. Other view-angle dependencies of detector performance, such as the additional blur due to oblique entry of x-rays, were found to be insignificant compared to other blur factors¹⁰ and were not included.

The logarithmic transformation was applied to the projection image, $S_p(f_{x'}, f_{y'})$ and $\Phi_p(f_{x'}, f_{y'})$ prior to FBP reconstruction. This process may be considered linear within a small exposure range and thus treated as a gain stage defined by $g=1/(KX)$. The resulting NPS and signal are given by¹⁴

$$\begin{aligned} S_I(f_{x'}, f_{y'}) &= \frac{S_p(f_{x'}, f_{y'})}{(KX)^2} \\ \Phi_I(f_{x'}, f_{y'}) &= \frac{\Phi_p(f_{x'}, f_{y'})}{KX}, \end{aligned} \quad (6)$$

where K is the x-ray sensitivity per pixel and X is the mean detector entrance exposure. This step produces a normalized NPS with the x-ray quantum noise component inversely proportional to radiation exposure.

Analogous to the reconstruction performed with the prototype system, the model employs a modified filtered backprojection algorithm.^{11,12} The RA and SA filter functions given in Eqs. (3) and (4) were applied to both signal and NPS after conversion to polar coordinates. A bilinear interpolation filter, $H_{IN}(f_r, f_y)$, was also applied to account for the voxel-driven reconstruction algorithm

$$H_{IN}(f_r, f_y) = \text{sinc}(\cos(\theta_i)m_x f_r) \text{sinc}(m_y f_y), \quad (7)$$

TABLE I. Summary of filters used in the FBP reconstruction and the frequency axes of their application.

Filter type	Direction of Application
Ramp (RA)	f_x, f_z
Spectral apodization (SA)	f_x
Slice-thickness (ST)	f_z

where m_x and m_y are the pixel dimensions of the projection image. The filtered signal and NPS, $S_f(f_r, f_y)$ and $\Phi_f(f_r, f_y)$, are given in polar coordinates

$$\begin{aligned} S_f(f_r, f_y) &= S_l(f_r, f_y) H_{RA}^2(f_r) H_{SA}^2(f_r) H_{IN}^2(f_r, f_y) \\ \Phi_f(f_r, f_y) &= \Phi_l(f_r, f_y) H_{RA}(f_r) H_{SA}(f_r) H_{IN}(f_r, f_y). \end{aligned} \quad (8)$$

$$\begin{aligned} S_b(f_x, f_y, f_z) &= \frac{N}{\theta_{TOT} f_r} \times \left[\sum_{i=1}^N S_f(f_r, f_y) \delta(f_x \sin(\theta_i) - f_z \cos(\theta_i)) \right] \times H_{ST}^2(f_z) \\ \Phi_b(f_x, f_y, f_z) &= \frac{N}{\theta_{TOT} f_r} \times \left[\sum_{i=1}^N \Phi_f(f_r, f_y) \delta(f_x \sin(\theta_i) - f_z \cos(\theta_i)) \right] \times H_{ST}(f_z) \end{aligned} \quad (9)$$

where f_x , f_y , and f_z are the Cartesian coordinates for the reconstructed spatial frequency domain and $\delta(f_x \sin(\theta_i) - f_z \cos(\theta_i))$ maps the signal and NPS of each projection along angle θ_i . The ST filter, H_{ST} , given by Eq. (5), was then applied in the f_z -direction to limit noise aliasing. At this stage, the variation in angular dose is reflected by the amplitude of the NPS along different angles. One example is shown in Fig. 1, where the central projections (limited by the inner angle, θ_{IN}) are acquired with higher dose than the outer projections (limited by the outer angle, θ_{TOT}). As a result, the amplitude of NPS in the central region I_{cent} is lower than that in the two peripheral I_{out} regions. When converting the sampling from polar to Cartesian coordinates, the NPS and signal must be normalized by the spoke density, as shown in Eq. (9). The presampling 3D MTF after backprojection (T_b) may be calculated by normalizing the output signal spectrum

$$T_b(f_x, f_y, f_z) = \Phi_b(f_x, f_y, f_z) / \Phi_b(0, 0, 0). \quad (10)$$

Because the proposed variable angular dose acquisition scheme does not affect the signal or NPS in the y -direction, the model was simplified to a 2D problem by setting $f_y = 0$. The response in the y -direction was assumed equivalent for all acquisition schemes.

II.B.2. Experimental implementation of variable angular dose distribution

Illustrated in Fig. 1 is our implementation of variable angular dose distribution in DBT acquisition. Several views around the central projection are acquired using a higher dose than that used for the peripheral projections. Angular exposures were distributed such that the total glandular dose is comparable to that of a standard DBT scan with uniform angular dose distribution.

The primary goal of our experimental study was to confirm the theoretical findings of the effect of angular dose distribution on the signal and noise properties of the reconstructed images. However, our prototype DBT system can only deliver dose uniformly at all angles by dividing the total exposure, specified in mAs, by the number of views. Our approach to variable angular dose distribution was to form a composite

The filtered signal and NPS are backprojected to the 3D spatial frequency domain using the central slice theorem. For a tomosynthesis acquisition with angular range θ_{TOT} and view number N , the output 3D NPS and signal $S_b(f_x, f_y, f_z)$ and $\Phi_b(f_x, f_y, f_z)$ may be calculated using

scan by selecting individual projection views from separate DBT scans acquired at different doses.

For each of our experiments, four DBT scans were acquired using the “ $\times 25$ ” mode at 28 kVp (W/Rh). The total exposures used for the four acquisitions were (1) 72, (2) 144, (3) 288, and (4) 432 mAs. The 144 mAs scan (5.76 mAs/view) was designated the reference scan, which corresponds to ~ 1.5 mGy glandular dose for a 4.2 cm breast of average density. To avoid detector saturation, 432 mAs was chosen as the maximum exposure setting. In conventional computed tomography (CT), the unattenuated x-ray intensity, I_0 , is measured. It is used to normalize the projection signal and to obtain the line integral of linear attenuation coefficients along the beam path. As a result, the reconstructed CT number is directly related to the linear attenuation coefficients of the object and independent of radiation dose. In our DBT system, the measurement of I_0 is unavailable due to detector saturation. Therefore, the logarithm of each projection is not normalized by I_0 prior to the application of reconstruction filter and backprojection.¹⁴ To avoid unequal weighting of each projection angle, the projection data at different ADS is scaled by their relative exposure ratio to the reference scan with uniform angular dose (i.e., scaled by 2 for each 72 mAs view, 1/2 for each 288 mAs view, and 1/3 for each 432 mAs view). As a result, the signal remains the same between different ADS and noise varies as a function of dose.

Two composite DBT scans were created by selecting projection views from the scaled DBT scans and are given in Table II, where the number of views, N_{views} , and exposures per view, X , are described for I_{cent} , I_{out-1} , and I_{out-2} regions. The reference scan with uniform angular dose distribution was designated angular dose scheme 1 (ADS-1). ADS-2 was composed by seven central views taken from the 288 mAs

TABLE II. Summary of acquisition schemes and respective exposures.

Scheme	$N_{views}(I_{cent})$	$X(I_{cent})$	$N_{views}(I_{out-1,2})$	$X(I_{out-1,2})$	X_{TOT}
ADS-1	0	—	25	5.76	144
ADS-2	7	11.52	18	2.88	132.48
ADS-3	5	17.28	20	2.88	144

scan (11.52 mAs/view) and 18 peripheral views taken from the 72 mAs scan (2.88 mAs/view). In ADS-3, the central five views were taken from the 432 mAs scan (17.28 mAs/view) and the peripheral 20 views were taken from the 72 mAs scan. The total glandular dose for both variable ADS was approximately equal to the reference (ADS-1) case, i.e., 1.5 mGy. Each ADS scheme was investigated for 3D noise characteristics as well as signal properties, i.e., modulation transfer function (MTF).

II.B.3. 3D NPS

The method for measuring 3D NPS has been described in detail previously.¹⁶ DBT scans were performed with a 3.95 cm thick Lucite block placed at the x-ray tube output. The reconstructed images are virtually scatter-free and contain only noise due to primary radiation. The volume of interest (VOI) with $512 \times 512 \times 40$ voxels at the center of the reconstructed volume was used for the 3D NPS calculation. The VOI was divided into 16 $128 \times 128 \times 40$ voxel subimages and the mean of each subimage was then subtracted and followed by a 3D Fourier transform. The NPS was calculated as the ensemble average of the 3D Fourier transform of each mean subtracted subimage

$$S(f_x, f_y, f_z) = \frac{d_x d_y d_z}{N_x N_y N_z} \langle |FT[I(x, y, z) - \bar{I}(x, y, z)]|^2 \rangle \quad (11)$$

where N_x , N_y , and N_z are the number of voxels in x , y , and z directions of each subimage I , and d_x , d_y , and d_z are the reconstructed voxel dimensions. The cross-sectional plane through $f_y = 0$ was used to visualize NPS (x - z), $S_{x-z}(f_x, f_z)$.

II.B.4. In-plane MTF

In principle the MTF of a DBT system should not change with dose. However, our experimental DBT system is implemented with continuous tube motion during x-ray exposure, which causes additional focal spot blur (FSB). To determine the extent of this effect in the context of variable angular dose acquisitions, the in-plane spatial resolution of the reconstructed images for each ADS was measured using the slanted edge method.^{17–19} The edge was made from a 200 μm thick sheet of 1100 Al alloy (99.0% pure) and placed 4 cm above the center of the chest wall side of the detector surface. The edge was aligned parallel to the detector surface and at a small angle (2 – 5°) from the y -direction (perpendicular to the direction of tube travel). The DBT scan of the slanted edge was then reconstructed and the oversampled edge spread function (ESF) was calculated from the image of the in-focus plane. The derivative of the ESF was calculated to generate the line spread function (LSF), the Fourier transform of which determines the in-plane MTF of the acquisition.

II.C. Investigation of object detectability as a function of ADS

The ideal observer SNR, d' , expressed in the frequency domain has been used extensively to evaluate the detectability of objects in projection radiography,²⁰ dual energy

imaging,^{21,22} and computed tomography (CT).^{23–25} To quantify the overall effect of ADS on object detectability, the ideal observer model was adapted for digital breast tomosynthesis and used as the figure-of-merit (FOM) for comparison.

II.C.1. Theory: Ideal observer signal-to-noise ratio for detection of Gaussian objects

The ideal observer SNR, or detectability index d' , is a dimensionless FOM for a signal embedded in a uniform background. For a three-dimensional (3D) image, d' is given by²⁶

$$d'_{3D^2} = \iiint \frac{K_c^2 [O^2(f_x, f_y, f_z) |T_b^2(f_x, f_y, f_z)|]}{S(f_x, f_y, f_z)} df_x df_y df_z, \quad (12)$$

where T_b is the 3D modulation transfer function, K_c is a multiplicative factor describing the radiographic contrast of the object, and O is the object function of the imaging task expressed in the frequency domain. In the present study, we used the detection task of a 3D Gaussian object

$$O(f_x, f_y, f_z) = e^{-\pi a_{\text{task}}^2 \sqrt{f_x^2 + f_y^2 + f_z^2}}, \quad (13)$$

where a_{task} defines its characteristic width.

Equation (12) describes the ideal observer's ability to detect an object within a 3D volume. Due to the anisotropic voxel size of DBT ($85 \mu\text{m} \times 85 \mu\text{m} \times 1 \text{ mm}$ in our system), the reconstructed images are exclusively viewed in the high resolution planes that are parallel to the detector (i.e., x - y plane). Thus, the in-plane detectability index was also calculated using²⁷

$$d'_{\text{in-plane}^2} = \iint \frac{[\int \Phi(f_x, f_y, f_z) df_z]^2}{\int S(f_x, f_y, f_z) df_z} df_x df_y \quad (14)$$

$$\Phi^2(f_x, f_y, f_z) = K_c^2 [O^2(f_x, f_y, f_z) |T_b^2(f_x, f_y, f_z)|]$$

Both the signal spectrum Φ and the NPS are integrated in the f_z -direction to yield the in-plane signal and NPS. Equation (14) uses the in-plane signal spectrum after DBT reconstruction to represent the 2D signal spectrum associated with a detection task. This is different from alternative approaches where the in-plane MTF is multiplied with a 2D task function to yield the signal spectrum.²⁸ Our approach takes into account the effect of limited resolution in z -direction, where the slice-thickness could be larger than the object size (e.g., calcifications). For these types of tasks, object detectability will depend on the 3D reconstruction, where the addition of ST filter is incorporated to minimize aliasing in the z -direction due to the coarse sampling distance of 1 mm.

Since the present study is focused on the relative performance between different angular dose schemes and the performance in the y -direction between the schemes are equivalent, the 3D problem was simplified to a 2D analysis in the x - z plane by setting $f_y = 0$. The in-plane detectability index calculation was therefore reduced to 1D using

$$d'_{2D, \text{in-plane}^2} = \int \frac{[\int \Phi(f_x, f_z) df_z]^2}{\int S(f_x, f_z) df_z} df_x. \quad (15)$$

FIG. 2. The x - z plane NPS for both the experimental measurements (left) and model (right). ADS-1 is shown in (a), ADS-2 is shown in (b), and ADS-3 is shown in (c). All schemes are plotted from -0.5 to 0.5 cycles/mm in the z -direction (vertical) and -5.88 to 5.88 cycles/mm in the x -direction (horizontal). These correspond to the Nyquist frequencies of each direction. An ST filter width $B = 0.1$ was used in all cases.

For the same reason of investigating relative performance between different ADS, the contrast factor K_c was also omitted and all d' calculations were normalized by the d' value of the uniform ADS-1 acquisition with a ST filter width of $B = 0.1$ under otherwise equivalent imaging conditions. The normalized d'_N is given by

$$d'_N = d' / d'(\text{ADS} - 1_{B=0.1}). \quad (16)$$

II.C.2. Phantom imaging experiment

To confirm qualitatively the dependence of object detectability on different ADS, three phantom imaging experiments were performed using the procedure outlined in Sec. II B 2. Images of the CIRS phantom were acquired and reconstructed for all ADS. The in-plane reconstructed images of the calcifications with sizes of 130 and 165 μm were compared for their relative detectability.

III. RESULTS AND DISCUSSION

III.A. Physical imaging performance using variable dose ADS

III.A.1. 3D NPS

Equation (6) was used to calculate the logarithmic transformed projection NPS at different doses and Eq. (9) was used to calculate the NPS after FBP reconstruction. All analyses are viewed for $f_y = 0$, since the NPS is not affected by ADS in the y -direction. Seen in Fig. 2 is the measured (left) and modeled (right) x - z plane NPS for ADS-1 (a), ADS-2 (b), and ADS-3 (c) with an ST filter width of $B = 0.1$. Compared with the uniform ADS-1, the variable ADS acquisitions exhibit lower NPS around the horizontal axis due to the high dose used in the central projections. The application of the ST filter eliminates the high-frequency components in the z -direction and reduces the noise contribution from the higher NPS in the peripheral views. The angular dependence of the NPS is observed consistently in both the measured and modeled results.

The in-plane NPS for different ADS was calculated by integrating the x - z plane NPS (shown in Fig. 2) in the z -direction. The results are shown in Fig. 3 as a function of f_x for four different reconstruction parameters. As shown in Fig. 3(a) for SBP reconstruction, ADS-1 with uniform angular dose distribution has the lowest in-plane NPS. This is because in order to keep the same total dose for variable

ADS acquisitions, a large number of peripheral projection views (18 for ADS-2 and 20 for ADS-3) were acquired with half of the dose of used in each view of ADS-1. As a result, the integrated in-plane NPS for ADS-3 is the highest, followed by ADS-2, due to the greater NPS contribution from the peripheral views. Shown in Figs. 3(b)–3(d) are results for FBP reconstruction in order of decreasing ST filter widths ($B = 0.1, 0.05$, and 0.035). Because the ST filter eliminates high-frequency components of the NPS in the peripheral projection views, the integrated in-plane NPS, shown in Fig. 3(b), exhibits a decrease in NPS at high spatial frequencies for variable ADS. Since the NPS at low spatial frequencies remains high, there exists a cross-over frequency point, f_c , at which the in-plane NPS for variable ADS drops below that of ADS-1. As ST filter width B decreases, as shown in Figs. 3(c) and 3(d), f_c for either ADS-2 or ADS-3 moves closer to the origin. This suggests better in-plane object detectability if the majority of the signal spectral power is above f_c (i.e., small, high contrast objects) and the signal spectrum does not vary with ADS (i.e., negligible FSB).

III.A.2. MTF

The in-plane presampling MTF measured with the thin Al edge are shown in Fig. 4 for all three ADS acquisitions. The results are in reasonable agreement with each other, suggesting a negligible effect of FSB between different ADS with the reconstruction filter parameters used in the present work ($A < 1.5, 0.035 < B < 0.1$). The discrepancy in measured in-plane MTF at low frequencies for each ADS may be attributed to the error in the derivation of ESF from experimental data due to the poor low-frequency response of in-plane images in DBT. The in-plane MTF measurements for $f > 1.5$ cycles/mm have excellent agreement between each ADS. Since microcalcifications express a larger proportion of their spectral power in higher frequencies, the effect of the difference in in-plane MTF due to FSB between each ADS is expected to be negligible. The in-plane NPS due to different ADS should be the dominant factor for d' .

III.B. Object detectability as a function of ADS

The 3D and in-plane d' were calculated using Eqs. (12) and (15), and the results normalized using Eq. (16). The presampling MTF were used in the calculation. In a previous

FIG. 3. Comparison of modeled 1D in-plane NPS in the x -direction (f_x) with $f_y = 0$ for all three ADS. The results in graphs (a)–(d) are for different reconstruction filter setting: (a) simple backprojection; (b) FBP with ST filter widths $B = 0.1$; (c) FBP with $B = 0.05$; and (d) FBP with $B = 0.035$.

investigation of frequency domain observer models using presampling MTF for projection mammography, the model result was found to agree well with the average performance of observers when the relative positions of the phantom and detector were shifted to vary the phase of the imaged objects.²⁹ Hence in the present investigation, the ideal observer SNR calculated using the presampling MTF of the detector can be assumed to be a phase averaged estimate of object detectability when aliasing is present in the system. It is important to note that the SA and ST reconstruction filters introduce additional blur in the f_x - and f_z -directions, respectively, and further reduce aliasing and phase dependence in the reconstructed images.

III.B.1. 3D Ideal observer detectability index, d'_{3D}

The 3D detectability index quantifies a system's ability to image and resolve an object over the entire volume, including the slice-thickness direction. The results for normalized 3D detectability d'_N are shown in Fig. 5 as a function of ST filter width B . The 3D MTF and task functions were assumed identical in all ADS, and only the 3D

FIG. 4. MTF for each acquisition scheme using a slanted Al edge phantom. The edge was placed 4 cm above the detector plane, and the MTF was evaluated at the in-plane location (the slice at which the edge resides in the reconstructed image).

NPS were varied in the calculation of d' . The graphs in Figs. 5(a)–5(d) correspond to d' of Gaussian objects with sizes of 1000, 500, 250, and 130 μm , respectively. Figure 5 shows a decrease in d' as B decreases. This is because the ST filter reduces the frequency response of the DBT system in the z -direction, i.e., increasing the effective slice-thickness. As a result, the 3D detectability d' decreases since the ability to resolve objects in the z -direction is compromised.

Compared with ADS-1, the drop in d' for ADS-2 and ADS-3 is slower as B decreases. This is because with variable ADS, the NPS in the peripheral views are higher than that in the central views, as shown in Fig. 1(b). These high NPS regions contribute less to the total 3D d' . Since the ST filter selectively removes the high-frequency components of the peripheral views [as may be seen in Fig. 1(b)], the loss in d' due to a decrease in B will be smaller for variable ADS. Since detection tasks for smaller objects have larger high-frequency components, this effect is more pronounced. As the object size decreases from 1000 to 130 μm in Figs. 5(a) to 5(d), the d' for variable ADS begins to outperform ADS-1 for any given B value.

III.B.2. In-plane detectability index

In order to calculate the in-plane d' , the in-plane signal spectrum was first obtained by integrating the 3D signal spectrum in the z -direction using Eq. (15). Figure 6 shows the in-plane signal power, which is the square of the in-plane signal spectrum, for different reconstruction filter settings and Gaussian object sizes. It shows that for the same object size, the application of a more aggressive ST filter causes a decrease in the in-plane signal power. Additionally, due to the selective removal of high-frequency components by the ST filter, the relative amplitude of the low-frequency components of the in-plane signal power increases.

Figure 7 shows the normalized in-plane d' for Gaussian objects of different sizes as a function of ST filter width B . Figures 7(a)–7(d) correspond to object sizes of 1000, 500, 250, and 130 μm , respectively. All plots are normalized to the d' value for ADS-1 with ST filter width of $B = 0.1$ using Eq. (16). Figure 7 shows that d' decreases with ST filter width, which is due to the reduced extent of the sampled frequency domain. As in the 3D d' case, ADS-2 and ADS-3 exhibit slower decreases in d' as ST filter width B decreases.

FIG. 5. Normalized 3D ideal observer detectability index, d'_N , for Gaussian objects as a function of slice-thickness filter width, B . Plots (a), (b), (c), and (d) correspond to Gaussian objects 1000, 500, 250, and 130 μm in size, respectively. ADS-1 is depicted in squares, ADS-2 in circles and ADS-3 in triangles.

Fig. 6. Comparison of modeled 1D in-plane signal power spectra in the x -direction (f_x), where $f_y = 0$ for objects of 1000, 500, 250, and 130 μm in size. The results in graphs (a)–(d) are for different reconstruction filter settings: (a) simple backprojection; (b) FBP with ST filter widths $B = 0.1$; (c) FBP with $B = 0.05$; and (d) FBP with $B = 0.035$.

The behavior of in-plane d' for different ADS can be better understood by comparing the changes in in-plane signal power (Fig. 6) and NPS (Fig. 3) for different object sizes and filter settings. Figure 3 shows that with more aggressive ST filters, the in-plane NPS exhibits a frequency point f_c , above which the NPS for ADS-2 and ADS-3 is lower than that for ADS-1. The value for f_c decreases with decreasing B . As shown in Fig. 6, the in-plane signal power is affected by both the object size and ST filter width. When there is substantial signal power above f_c , variable ADS become beneficial. This corresponds to the smallest object size in Fig. 7(d) with ST filter width of $B < 0.1$.

Both 3D and 2D d' results show that the ideal observer SNR for the detection of small objects with different ADS depends strongly on the imaging task and the reconstruction filter settings. The detection of larger objects (e.g., 1 mm calcification in a uniform, stochastic noise background) favors

uniform ADS. For imaging tasks with substantial high-frequency signal power, e.g., the detection of objects smaller than 200 μm , variable ADS becomes beneficial when a ST filter is applied to minimize aliasing in reconstructed slices (at the cost of resolution in the z -direction). Future studies using other imaging tasks with higher spatial frequency components, e.g., discrimination tasks, or different observer models, may lead to different optimization strategies.

Under the imaging conditions where variable ADS is advantageous, some other system performance factors need to be taken into consideration. The maximum dose to the central projections is limited by the total dose of the entire scan and the dose to the peripheral views should be above the quantum noise limited level dictated by the detector electronic noise. As dose is more concentrated on central projections, more aggressive ST filters are required to exhibit benefit over ADS-1 even for small objects. Also, as the

FIG. 7. Ideal observer normalized detectability index, d_N , for Gaussian objects as a function of slice-thickness filter width, B . Plots (a), (b), (c), and (d) correspond to Gaussian objects 1000, 500, 250, and 130 μm in size, respectively. ADS-1 is depicted in squares, ADS-2 in circles, and ADS-3 in triangles.

value of B decreases, the resolution in z -direction is reduced, limiting the ability of DBT to remove out-of-plane structures. This would degrade the detection of masses, which is dominated by structural noise. In addition, it has been shown by Das *et al.* that reconstructed images using all projection views provide better detection of microcalcifications than using the central projection (with half the dose) alone.⁹ The wider impact of the present work is to provide a cascaded linear-system model to predict the imaging performance of DBT as a function of angular dose distribution, so that the best imaging strategy may be chosen for a given task and a specific set of imaging conditions.

III.B.3. CIRS phantom

Figure 8 shows the in-plane reconstructed images of the calcifications from the CIRS tissue-equivalent mammography phantom. The region of interest (ROI) includes the 130 μm (top) and 165 μm (bottom) calcifications. All images were adjusted for equivalent contrast. The theoretical results in the

previous section were used to select appropriate reconstruction filter settings: SA filter width of $A = 1.5$ and ST filter width of $B = 0.035$. Figure 8 shows an improved conspicuity with in-plane images of small calcifications for both variable ADS schemes. A qualitative review of the noise behavior of the images also reveals a decrease in apparent quantum noise for both variable ADS acquisitions over ADS-1, which is in agreement with the modeled results in Fig. 3(d).

III.C. Discussion: Assumptions used in our cascaded linear-system model

In this section, we discuss the validity and impact of several assumptions used in our cascaded linear-system model.

III.C.1. Scattered radiation

Scattered radiation is not included in our present implementation of the cascaded linear-system model. It has been shown in previous studies by Kyrianiou *et al.* that the main

FIG. 8. In-plane images of the 130 μm (top) and 165 μm (bottom) calcifications in the CIRS 011A tissue-equivalent mammography phantom. An SA filter width of $A=1.5$ and an ST filter width of $B=0.035$ were used in all cases and all images were adjusted for equivalent contrast.

effect of scatter is a low-frequency (<0.1 cycles/mm) drop in MTF.³⁰ Its effect on the detection of small microcalcifications, which retains its spectral power predominantly in the high frequencies, should be negligible. The scatter to primary ratio (SPR) increases with projection angle due to the increase in effective breast thickness. The SPR for a 5 cm breast has been shown to increase by $\sim 8\%$ with an oblique entry of 18° .³¹ However, since the SPR at a given angle is independent of dose, the relative increase in NPS due to scattered radiation and its dependence on projection angle is the same for each ADS. Our results compare the relative performance between ADS rather than the absolute SNR. The effect of scatter should therefore be negligible. However, future investigation of absolute SNR and object detectability should incorporate the effect of scatter and its dependence on projection angle.

III.C.2. Focal spot blur due to continuous tube motion

Our analysis was generalized for a system without FSB. This assumption applies to the following conditions:

FIG. 9. The modeled in-plane MTF for each ADS. Each modeled MTF includes the effect of FSB due to focal spot motion in our prototype system.

(1) DBT system with step-and-shoot gantry motion, i.e., no focal spot motion during exposure and (2) a system where the effect of FSB is negligible compared to other image blur factors, e.g., detector MTF, pixel binning, and reconstruction filters. For a 25 view, full resolution acquisition using our prototype system, the MTF due to FSB is >0.5 at f_{NY} for exposures less than 14.2 mAs per view, which corresponds to ADS-1 scans of up to 355 mAs.

While FSB due to continuous tube motion may degrade the projection MTF in the central projections, it remains negligible in peripheral views. As a result, the in-plane MTF is essentially the same for all ADS. Figure 9 shows the modeled in-plane presampling MTF for all three ADS, which includes the effect of FSB due to focal spot motion. They are virtually identical.

It is important to note that FSB does not cause correlation in the NPS, hence the overall effect of FSB on d' should be negligible. With our prototype system in full resolution mode, the FSB causes a $\sim 2.36\%$ drop in $d'_{\text{in-plane}}$ with ADS-3 for a 130 μm Gaussian object ($A=1.5$ and $B=0.035$). Therefore, our assumption of no FSB is reasonable for the purpose of d' comparison.

IV. SUMMARY

The effect of angular dose distribution in DBT acquisition was investigated. Our results show that delivering more dose toward the central views at the expense of reduced dose and increased noise in the peripheral views results in major changes in the 3D and in-plane NPS. Its impact on ideal observer detectability index, d' , depends greatly on the width of the ST filter applied in the z -direction, which limits the extent of the frequency domain used for reconstructed images. For clinically relevant ST filter widths ($B=0.035$ for full resolution images), there may be some benefit for implementing a variable ADS when the task involves the detection of very small objects.

The cascaded linear-system model provides a useful tool to determine the effect of angular dose distribution on 3D NPS in DBT. In the present work, the modeled 3D signal and NPS was used to calculate the ideal observer d' for the detection of Gaussian objects embedded in a uniform background under different ADS and reconstruction filter settings. The results showed that both 3D and 2D d' decrease with ST filter width. However with clinically relevant B filter settings, which is used to minimize noise aliasing, variable ADS could provide higher d' than uniform ADS for the detection of small objects (e.g., $<200 \mu\text{m}$). A more generalized analysis including the effect of breast structural noise on the detection of both masses and calcifications is currently under investigation and not within the scope of the present work.

ACKNOWLEDGMENTS

The authors gratefully acknowledge the financial support from NIH (1 R01 CA148053 and 1 R01 EB002655) and Siemens Healthcare. The authors thank Dr. Thomas Mertelmeier and Dr. Jasmina Ludwig for helpful discussion and assistance. The authors would also like to thank the

anonymous reviewers for their helpful comments and suggestions to improve our manuscript.

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