Experimental validation of a three-dimensional linear system model for breast tomosynthesis

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A three-dimensional (3D) linear model for digital breast tomosynthesis (DBT) was developed to investigate the effects of different imaging system parameters on the reconstructed image quality. In the present work, experimental validation of the model was performed on a prototype DBT system equipped with an amorphous selenium (a-Se) digital mammography detector and filtered back-projection (FBP) reconstruction methods. The detector can be operated in either full resolution with $85\text{ }\mu\text{m}$ pixel size or 2×1 pixel binning mode to reduce acquisition time. Twenty-five projection images were acquired with a nominal angular range of $\pm 20^\circ$. The images were reconstructed using a slice thickness of 1 mm with 0.085×0.085 mm in-plane pixel dimension. The imaging performance was characterized by spatial frequency-dependent parameters including a 3D noise power spectrum (NPS) and in-plane modulation transfer function (MTF). Scatter-free uniform x-ray images were acquired at four different exposure levels for noise analysis. An aluminum (Al) edge phantom with 0.2 mm thickness was imaged to measure the in-plane presampling MTF. The measured in-plane MTF and 3D NPS were both in good agreement with the model. The dependence of DBT image quality on reconstruction filters was investigated. It was found that the slice thickness (ST) filter, a Hanning window to limit the high-frequency components in the slice thickness direction, reduces noise aliasing and improves 3D DQE. An ACR phantom was imaged to investigate the effects of angular range and detector operational modes on reconstructed image quality. It was found that increasing the angular range improves the MTF at low frequencies, resulting in better detection of large-area, low-contrast mass lesions in the phantom. There is a trade-off between noise and resolution for pixel binning and full resolution modes, and the choice of detector mode will depend on radiation dose and the targeted lesion. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3040178](https://doi.org/10.1118/1.3040178)]

Key words: breast tomosynthesis, amorphous selenium detector, pixel binning, MTF, NPS

I. INTRODUCTION

Projection mammography is the standard method for breast screening; however, its sensitivity and specificity for cancer detection is limited due to “structural noise” caused by breast tissue overlap. In digital breast tomosynthesis (DBT), several projection images of the breast are acquired at different angles. This is generally accomplished by rotating the x-ray tube around a pivot point near the center of the breast. Depending on design, the detector may be stationary or move in synchrony with the x-ray tube.¹ The thickness of each reconstructed slice is typically 1 mm, thus reducing significantly the effect of overlapping breast tissue. Promising clinical results have prompted intensive investigation of this three-dimensional (3D) imaging technique for both screening and diagnosis.^{1–4}

The reconstructed image quality in DBT depends on many factors. To achieve the best clinical outcome it is cru-

cial to understand the fundamental imaging characteristics of this technology. Resolution and noise transfer characteristics are standard metrics for image quality assessment in both 2D radiography and 3D tomography. Both the modulation transfer function (MTF) and noise power spectrum (NPS) have been used as important parameters for detector characterization in projection imaging.^{5,6} These image quality metrics have also been extended to 3D for conventional computed tomography (CT) or cone beam CT (CBCT). In CT the MTF was usually characterized using the point spread function (PSF) derived from the cross-sectional image of a wire placed in the longitudinal direction of a CT scanner.⁷ Methods for measuring the presampling MTF were also developed using the slanted edge (for CT),⁸ or a wire⁹ and sphere¹⁰ (for CBCT). The theoretical model and experimental measurement of 3D NPS were conducted for both CT (Ref. 11) and CBCT.¹² It was found that the NPS is asymmetrical and dependent on reconstruction filters.^{11,12}

FIG. 1. Simplified diagram showing the 3D cascaded linear system model for our DBT system.

Quantitative measurements of image quality for DBT, in terms of resolution and noise, have been performed mainly using in-plane images, i.e., the plane parallel to the detector. Chen *et al.* measured the in-plane NPS and noise equivalent quanta (NEQ) with scattered radiation and three reconstruction methods.¹³ The in-plane spatial resolution was studied by different investigators through simulated images of a point^{14,15} and a wire,^{16,17} or experimental measurements using edge^{18,19} and wire²⁰ phantoms. Previously we built a computer simulation platform to study the effect of imaging geometry and reconstruction methods.¹⁶ The in-plane presampling MTF was calculated from the reconstructed images of a slanted tungsten (W) wire. It was found that the in-plane MTF was determined by a reconstruction filter as well as imaging geometry. The determination of resolution in the thickness direction has been quantified by the artifact spread function (ASF) (Refs. 16 and 21–23) and slice sensitivity profile (SSP) (Refs. 19 and 24) in the spatial domain. The quality of the phantom^{4,22} and patient images^{3,21,25} was investigated as a function of reconstruction algorithms and imaging conditions at a given glandular dose. Due to the limited view number and angular range of DBT acquisition, previous investigations were mainly focused on the effects of reconstruction methods. There is a lack of comprehensive measurement on the 3D noise characteristics of DBT.

In a companion paper, we developed a cascaded linear system model for DBT using similar methods as that developed for CT and CBCT. Using this systematic approach, the effects of imaging geometry, reconstruction filter, detector performance, and radiation dose on 3D imaging performance metrics can be integrated into the same analysis framework. In the present article, the experimental measurements of DBT imaging metrics were performed using a prototype system. The measured 3D NPS and in-plane MTF were compared with results from our theoretical model. An ACR phantom was also imaged to investigate the impact of acquisition geometry and reconstruction filters.

II. CASCADED LINEAR SYSTEM THEORY

The details of the 3D cascaded linear model developed for DBT were described in the companion paper.²⁶ The model was implemented with the detector performance, reconstruction filter, imaging geometry, and x-ray exposure that correspond to the tomosynthesis prototype system used in the present work. Here, we will only provide a brief outline of the model framework. Figure 1 shows the flow chart of the model. It begins with the 2D signal (Φ_p) and noise power spectra (S_p) of the projection images calculated using a cascaded linear system model previously developed for amorphous selenium (a-Se) flat-panel detectors.⁶ The additional

FIG. 2. Schematic diagram showing the geometry of the prototype DBT system: The x -axis is the tube travel direction, the z -axis is along the thickness of the breast, and the y -axis is orthogonal to x - z plane (not shown in the figure).

focal spot blur due to x-ray tube motion in DBT was also included. Then, Φ_p and S_p were modified by the reconstruction filters. The filtered signal and NPS at different projection angles θ_i , Φ_h , and S_h , were mapped into the 3D space using the central slice theorem.²⁷ The 3D spectra, Φ_b and S_b , were aliased at multiples of 2 times the Nyquist frequency, f_{NY} , in each direction to form the final 3D spectra, Φ_v and S_v , from which the 3D MTF, NPS, and DQE could be derived. As with all cascaded linear system analysis, the model assumes shift invariance, which is only possible with a parallel beam. Given the minimal magnification in DBT (because the breast is compressed on top of the detector surface), it is a reasonable assumption.

III. METHOD

III.A. Tomosynthesis prototype system

Experimental measurements were performed on a prototype DBT system equipped with an a-Se flat-panel full field digital mammography (FFDM) detector. (Breast tomosynthesis is an investigational practice and is limited by U.S. law to investigational use. It is not commercially available in the U.S. and its future availability cannot be ensured). The system was modified from a Siemens Novation screening mammography unit with target/filter combinations of molybdenum(Mo)/Mo, Mo/rhodium(Rh), and W/Rh.²⁸ As shown in Fig. 2, the breast is compressed on top of a stationary a-Se detector with 85 μm pixel size. The center of rotation (COR) of the tube column is 6.0 cm above the detector surface, and the source to imager distance (SID) is 65 cm, the same as that in screening mammography. The detector can be read out with full resolution or 2×1 pixel binning in the tube travel direction (x -axis, Fig. 2) to increase the read-out rate. In the present study, the target/filter combination of

TABLE I. Image acquisition and reconstruction parameters for the two DBT imaging modes used in our experimental investigation.

Imaging mode		x25	x25bin
Acquisition parameters	Detector operation mode	Full resolution	2×1 binning
	Angular range	$\pm 20^\circ$	$\pm 20^\circ$
	View number	25	25
	Scanning time	20.8s	12.7s
	Total mAs	144 mAs	144 mAs
	Focal spot motion per view	0.65 mm	1.15 mm
Reconstruction parameters	Softbinning(2×2)	No	Yes
	Nyquist frequency (f_{x-NY})	5.88 cycles/mm	2.94 cycles/mm
	A (SA window width)	1.5 (8.8 cycles/mm)	1.5 (4.4 cycles/mm)
	B (ST window width)	0.1 (0.59 cycles/mm)	0.2 (0.59 cycles/mm)

W/Rh with 28 kVp was used exclusively. The half-value layer (HVL) of this spectrum was measured as 0.53 mm (Al).

The images can be acquired in several different modes with a tube travel angular range of $\pm 25^\circ$ and the number of views ranging from 11 to 49. The nominal angular range for image acquisition, defined as the angular separation between the first and last views, is $\pm 20^\circ$. Two imaging modes were used in the present investigation, and their image acquisition and reconstruction parameters are summarized in Table I. In mode x25, 25 views were acquired with full detector resolution within 20.8 s. In mode x25bin, 25 views were acquired in detector binning (2×1) mode, which reduces the acquisition time to 12.7 s. In this mode, a software-binning in the y -direction (perpendicular to the x -direction, Fig. 2), was applied on the projection image before reconstruction to form an effective pixel size of $170 \mu\text{m}$. One consequence of reduced scan time is the increase in focal spot blur (FSB) due to a larger tube travel distance during x-ray exposure.²⁹ The effects of degraded spatial resolution on the detection of high-frequency information, e.g., calcification, will be investigated.

The image reconstruction was performed with a filtered backprojection (FBP) algorithm described in detail previously,³⁰ and only a brief outline will be provided here. The filters included three components: (1) ramp (RA) filter; (2) spectrum apodization (SA) filter to limit high-frequency noise; and (3) slice thickness (ST) filter, which is applied in the z -direction (perpendicular to the detector plane, Fig. 2). Both the SA and ST filters use the Hanning window. The width of the Hanning window is given in multiples of the Nyquist frequency, f_{x-NY} , of the projection images. The window width multiplicative factors for the SA and ST filters are denoted A and B , respectively. Larger A and B values lead to a higher frequency response, with the penalty of higher noise in the reconstructed images. In the present work $A=1.5$ was chosen, resulting in a Hanning window function that is ~ 0.25 at the Nyquist frequency of the projection images. $B=0.1$ and 0.2 were chosen for the full resolution and binning mode, respectively. They led to a Hanning window function of ~ 0.05 at the Nyquist frequency in the z -direction, $f_{z-NY}=0.5$ cycles/mm. This ensures minimal noise aliasing in the z -direction while preserving the high-

frequency information in the projection images. The A and B values for the two imaging modes are listed in Table I. To investigate the effects of image reconstruction on DBT image quality, four different filter schemes were applied for each acquisition mode, and the details are summarized in Table II.

Since most of the images in our present investigation were uniform noise images or for small phantoms, a small reconstructed volume of interest (VOI) containing 40 slices was used for all images in our investigations. The number of in-plane pixels varied from 512×512 (for noise and MTF measurement) to 1000×1000 (phantom study). The projection image for each view was cropped according to the VOI prior to the application of filters to minimize reconstruction time. The slice thickness was 1 mm, and the in-plane pixel size was $0.085 \times 0.085 \text{ mm}^2$, resulting in an anisotropic voxel size of $0.085 \times 0.085 \times 1 \text{ mm}^3$. The same voxel size was used for both acquisition modes to facilitate comparison despite their difference in projection image resolution.

III.B. 3D NPS

The experimental method for 3D NPS measurement is depicted in Fig. 3. To avoid the effect of scattered radiation on the NPS measurements, a 3.95 cm thick Lucite block was placed at the x-ray tube output during image acquisition. The uniform noise images were acquired using both x25 and x25bin modes at four different exposure levels ranging from 72 to 576 mAs, corresponding to a detector entrance exposure of 0.59–4.71 mR per view. As a reference for comparison, the mean glandular dose to a 4.2 cm breast with average density (50/50) is 1.5 mGy with 144 mAs (513 mR) of exposure. It has been shown in Ref. 29 that the effect of detector electronic noise on DQE becomes negligible for detector entrance exposures of >1 mR. At 0.4 mR, the electronic noise degrades the DQE by $\sim 10\%$. Therefore, we could ig-

TABLE II. Summary of the filter schemes for FBP reconstruction.

Filter scheme 1	Simple backprojection (SBP)
Filter scheme 2	H_{RA} only
Filter scheme 3	$H_{RA} \times H_{SA}$
Filter scheme 4	$H_{RA} \times H_{SA} \times H_{ST}$

FIG. 3. The imaging geometry for NPS measurement is shown on the left. An example of a projection image with uniform noise is shown in the lower right, and the dimensions of the reconstructed volume are shown in the upper right.

nore the effect of detector noise, and focus our investigation on the other factors such as imaging geometry and reconstruction filters.

The reconstructed 3D images contained only noise, and the VOI of $512 \times 512 \times 40$ voxels was divided into 16 sub-images, each containing $128 \times 128 \times 40$ voxels. One realization of the 3D NPS was calculated from each subimage by taking the 3D Fourier transform of the mean subtracted volume data. The final NPS, $S(f_x, f_y, f_z)$, was obtained as the ensemble average of all the NPS realizations,

$$S(f_x, f_y, f_z) = \frac{d_x d_y d_z}{N_x N_y N_z} \langle |FT[I(x, y, z) - \bar{I}(x, y, z)]|^2 \rangle, \quad (1)$$

where N_x , N_y , and N_z are the number of voxels in the x , y , and z directions, respectively, for each subimage, d_x , d_y , and d_z are the voxel dimension in the corresponding directions (Fig. 3), and $I(x, y, z)$ is the reconstructed image intensity. The voxel noise variance (σ_v^2) in the spatial domain could be determined as the integral of the 3D NPS as

$$\sigma_v^2 = \int \int \int S(f_x, f_y, f_z) df_x df_y df_z. \quad (2)$$

After the 3D NPS was calculated, an evaluation of the in-plane 2D NPS for any selected slice, $S_{\text{avg}}(f_x, f_y)$, could be obtained by integrating the 3D NPS along the z -direction up to $f_{z-\text{NY}}$,

$$S_{\text{avg}}(f_x, f_y) = \int_{-f_{z-\text{NY}}}^{f_{z-\text{NY}}} S(f_x, f_y, f_z) df_z. \quad (3)$$

In a stationary system, $S_{\text{avg}}(f_x, f_y)$ from Eq. (3) should be equal to the 2D NPS calculated using any slice of the reconstructed image. However, since our prototype system uses a cone beam, it is not strictly stationary. To test the assumption of stationarity in our linear system model, we compared the $S_{\text{avg}}(f_x, f_y)$ obtained using Eq. (3) with the 2D NPS calculated using individual reconstructed slices. In our tomosynthesis geometry with limited angular range, the top slice ($z = 40$) is expected to deviate the furthest from a parallel beam

FIG. 4. Diagrams showing the experimental method for in-plane MTF measurement: Imaging geometry with the edge phantom elevated from the detector surface by a 4 cm PMMA block (left); an example of a reconstructed in-plane edge image (center); and the ESF function derived from the edge image (right).

assumption. Therefore, its NPS (S_{40}) was used for comparison, and the values were calculated using the standard method for 2D projection images,

$$S_{40}(f_x, f_y) = \frac{d_x d_y}{N_x N_y} \langle |FT[I(x, y, z = 40) - \bar{I}(x, y, z = 40)]|^2 \rangle. \quad (4)$$

III.C. Spatial resolution

The spatial resolution of the prototype DBT system was characterized using its in-plane presampling MTF, which was measured using the slanted edge method.^{31–33} An aluminum (Al) edge phantom with 0.2 mm thickness was made using a 99% pure Al sheet. This particular thickness was chosen for the following reasons: (1) The shadow created by the thickness of the edge at the maximum tube angle (20°) should not introduce any error in the MTF measurement; (2) the projection radiographic contrast of the edge phantom should be reasonably small so that the logarithmic transformation before FBP reconstruction can be regarded as a linear process.

As shown in Fig. 4, the edge was placed 4 cm above the center of the detector surface near the chest wall side, and at a small angle ($2-5^\circ$) from the gate line (y -) direction of the detector. The projection images were acquired using x25 mode with a total exposure of 144 mAs, corresponding to a total raw detector exposure of 523 mR.

The measurement of the in-plane MTF was performed using the reconstructed image slice for the plane where the edge was placed. It has the highest image contrast for the edge. Figure 4 shows an example of the reconstructed edge image, from which the edge spread function (ESF) was deduced. As shown in Fig. 4, the ESF shows a strong edge enhancement, which is characteristic of FBP reconstruction with a limited angular range.

III.D. Phantom imaging experiment

The above frequency-domain image quality metrics provides quantitative evaluation of the imaging performance, and facilitates the fundamental understanding of the effects of system parameters on DBT. In this section we investigate the imaging characteristics of different objects after DBT reconstruction, and relate the results to spatial frequency domain image quality metrics. An ACR mammography accreditation phantom (Nuclear Associates) was imaged with different acquisition modes and reconstruction filters. Two angular spans were used: (1) $\pm 20^\circ$; and (2) $\pm 10^\circ$ with the same angular separation and total glandular dose (1.5 mGy). The second configuration was achieved by doubling the total mAs for the DBT scan and reconstructing the 3D image using half (13) of the acquired images. Since two detector operational modes can be used for each angular span, a total of four DBT imaging modes were resulted: (1) binning mode with $\pm 20^\circ$; (2) binning mode with $\pm 10^\circ$, (3) full resolution with $\pm 20^\circ$; and (4) full resolution with $\pm 10^\circ$. The reconstructed image slice for the plane that contains the wax insert of the ACR phantom, where the images of all the objects are in sharp focus, was used for image quality evaluation. For comparison, the ACR phantom was also imaged using the screening mode on the prototype DBT system, i.e., one exposure with the tube column at a zero-degree angle. The image was acquired with the same x-ray spectrum and total exposure (140 mAs) as that used in DBT.

III.D.1. Calcifications

The reconstructed image intensity profile of the largest calcification specks, i.e., cal-1 with diameter $d=0.54$ mm, was examined to investigate the effect of image acquisition and reconstruction parameters on the detection of small and high contrast object. The profile was taken across the center of cal-1 and normalized by the maximum intensity. This method facilitates quantitative comparison between the contrast and sharpness of the detection of calcifications.

III.D.2. Mass

The reconstructed image quality of mass was quantified using the signal difference to noise ratio (SDNR). The SDNR for the two largest masses: mass-1 with $d=2$ mm and mass-2 with $d=1$ mm, was calculated using

$$\text{SDNR} = (\rho_{\text{mass}} - \rho_{\text{bak}}) / \sigma_{\text{bak}}, \quad (5)$$

where ρ_{mass} and ρ_{bak} are the mean reconstructed in-plane pixel intensity values for the mass and the background, respectively, and σ_{bak} is the standard deviation of the background. The value for ρ_{mass} was computed from a circular region of interest (ROI) with a 50 pixel diameter in the center of the mass. The background was chosen as an ROI with a 100 pixel diameter in the center of the phantom where no object was present.

TABLE III. Voxel variance as a function of total mAs with reconstruction scheme 3 in full resolution (x25) and binning (x25bin) mode. The variance vs exposure curves were fitted with a reciprocal relationship.

Total mAs	Voxel variance (x25)	Voxel variance (x25bin)
72	2.64×10^{-3}	7.09×10^{-4}
144	1.13×10^{-3}	2.87×10^{-4}
288	5.33×10^{-4}	1.43×10^{-4}
576	2.93×10^{-4}	7.87×10^{-5}

The image quality of each object was investigated as a function of the following two DBT parameters: (a) filter schemes as listed in Table II; and (b) angular range and detector operation mode as listed in Table I.

IV. RESULTS AND DISCUSSION

IV.A. NPS

For easy comparison and visualization of the results, the 3D NPS measurements are evaluated using three methods: (1) the integrated voxel variance to quantify the overall perception of noise in the reconstructed images; (2) in-depth NPS, which is the 3D NPS in the x - z plane to visualize the effect of reconstruction filters and angular range; and (3) in-plane NPS, which is directly related to the slice-by-slice viewing of reconstructed DBT images.

IV.A.1. Voxel variance

Shown in Table III are the voxel variance values calculated using Eq. (2) as a function of total mAs. Filter scheme 3 was used for both x25bin and x25 modes. Table III shows that the voxel variance is inversely proportional to exposure. This is due to the gain of logarithmic transformation of the projection images before FBP.²⁶ Because the detector performance is dominated by x-ray quantum noise at the exposure level used in each scan,²⁹ the NPS of the projection image after logarithm transformation is inversely proportional to the mean of the projection image, and hence to the entrance exposure. Table III also shows that the variance in full resolution mode is approximately four times that in binning mode. This is consistent with the high intrinsic resolution of a-Se detectors, which leads to a quadrupled increase in the number of x-ray quanta detected per pixel after 2×2 binning (hardware binning in the x -direction and software binning in the y -direction).

IV.A.2. In-depth NPS

Figure 5 shows the comparison between measured and modeled in-depth (x - z) NPS (with $f_y=0$) for x25 mode using all four filter schemes. For ease of comparison, the modeled results are scaled such that their total integrated NPS match that of the experimental measurements. The total exposure used for the acquisition was 144 mAs. The center of each NPS image is the zero frequency, and the results are shown up to f_{NY} in each direction ($f_{x\text{-NY}} = \pm 5.88$ cycles/mm and $f_{z\text{-NY}} = \pm 0.5$ cycles/mm). The modeled and measured NPS

FIG. 5. Comparison between modeled (top) and measured (bottom) NPS for mode x25 using all four reconstruction schemes listed in Table II. (a), (b), (c), and (d) are for filter schemes 1, 2, 3, and 4, respectively. The total exposure used for the acquisition was 144 mAs. The NPS is shown up to f_{NY} in each direction ($f_{x-NY} = \pm 5.88$ cycles/mm and $f_{z-NY} = \pm 0.5$ cycles/mm).

results show an essentially identical pattern, which proves that the central slice theorem can be used in the linear system model to fill data in the 3D frequency space. Near the origin (zero frequency), only a small angular range of the frequency space is filled, which is consistent with the angular range ($\theta = \pm 20^\circ$) used in our DBT acquisition. Figure 5(a) shows that the 3D NPS with SBP is simply the projection NPS replicated at different angles; therefore, the intensity of NPS is the highest at the origin. Noise aliasing is manifested as increased NPS intensity at high frequencies due to the replication of NPS at multiples of $2f_{NY}$. With the application of the RA filter, as shown in Fig. 5(b), the NPS decreases at low frequencies and increases at high frequencies, causing the noise aliasing to be more visible. The application of an SA filter [Fig. 5(c)] reduces the magnitude of high-frequency noise. The main difference with the application of ST filter [Fig. 5(d)] is that it essentially eliminated the frequency components above f_{z-NY} ; hence, noise aliasing is negligible.

Shown in Fig. 6 are the calculated and measured in-depth NPS for x25bin mode. The main difference, compared to the results in Fig. 5, is the reduced frequency response in the x -direction due to pixel binning, which formed an effective pixel size of $170 \mu\text{m}$. Since the voxel dimension of the re-

constructed images is the same as in full resolution ($0.085 \times 0.085 \times 1 \text{ mm}^3$), the f_{NY} of the reconstructed images is unchanged. Although binning helps reduce high-frequency noise, the degraded frequency response affects the detection of different objects. Qualitative evaluation for this effect will be addressed in the phantom study.

To demonstrate the effect of angular range, the noise images were reconstructed using a subset of the projection images with an angular span of $\pm 10^\circ$. The resulting in-depth NPS are shown in Fig. 7 for reconstruction schemes 3 and 4, and the results for the full angular range (Fig. 5) are shown for comparison. The total dose for the two angular ranges was comparable. Figure 7(a) shows that the sampled area at low frequencies decreases with a decrease in angular span. At high frequencies, the noise aliasing with a wider angular span is more severe. After the ST filter is applied, as shown in Fig. 7(b), the aliasing at high frequencies is eliminated. The limited sampling at low frequencies for a smaller angular range is unchanged. The angular range will affect the detection of low-frequency objects, i.e., masses, as will be discussed in the phantom study (Sec. III C).

FIG. 6. Comparison between modeled (top) and measured (bottom) NPS for mode x25bin using all four reconstruction schemes listed in Table II. (a), (b), (c), and (d) are for filter schemes 1, 2, 3, and 4, respectively. The total exposure used for the acquisition was 144 mAs. The NPS is shown up to f_{NY} in each direction ($f_{x-NY} = \pm 5.88$ cycles/mm and $f_{z-NY} = \pm 0.5$ cycles/mm).

FIG. 7. Measured in-depth NPS with two angular ranges: $\pm 20^\circ$ (top) and $\pm 10^\circ$ (bottom) for (a) filter scheme 3 and (b) filter scheme 4. The detector was operated in full resolution mode, and the total mAs used for each scan was ~ 144 mAs. The NPS is shown up to f_{NY} in each direction ($f_{x-NY} = \pm 5.88$ cycles/mm and $f_{z-NY} = \pm 0.5$ cycles/mm).

IV.A.3. In-plane NPS

Shown in Figs. 8(a) and 8(b) are the modeled and measured in-plane NPS for modes x25 and x25bin, respectively. The top row of each figure shows the modeled results for all four filter schemes. Shown in the middle row are the experimental results derived from the 3D NPS measurement using Eq. (3), which provides an estimate of the average in-plane NPS of all slices. Shown in the bottom row are the 2D NPS of the top slice. The in-plane 2-D NPS are shown up to $f_{x-NY} = f_{y-NY} = \pm 5.88$ cycles/mm. The measured NPS con-

tains higher noise on the zero frequency axes, which is caused by the nonuniformity in the 3D images.

Figure 8 shows that, except for filter scheme 1 with SBP reconstruction, all FBP schemes have anisotropic in-plane NPS. The decrease in NPS in the y-direction is due to the interpolation filter used in our voxel-driven FBP reconstruction. The NPS at high frequencies is lower for x25bin mode due to pixel binning in both directions before FBP reconstruction.

To facilitate quantitative comparison, the measured and modeled NPS in the x-direction at $f_y = 0$ cycles/mm, i.e., the central slice of the 2D NPS in Fig. 8, are plotted in Fig. 9. Due to the high noise on central axes in the measured data,

FIG. 8. (a) In-plane NPS for x25 mode with all four reconstruction schemes (column 1–4 from the left for filter schemes 1–4) The top row is the in-plane NPS calculated using the model; the middle row is the average in-plane NPS from the measurement; and the bottom row is the measured NPS of the top slice. (b) In-plane NPS for x25bin mode with all four reconstruction schemes (column 1–4 from the left for filter schemes 1–4) The top row is the in-plane NPS calculated using the model; the middle row is the average in-plane NPS from the measurement; and the bottom row is the measured NPS of the top slice.

FIG. 9. (a) Comparison between modeled and measured in-plane NPS in the x-direction (with $f_y = 0$ cycles/mm) for filter schemes 3 and 4 in mode x25. (b) Comparison between modeled and measured in-plane NPS in the x-direction for filter schemes 2–4 in mode x25bin.

FIG. 10. (a) Measured in-plane presampling MTF with the four reconstruction schemes for acquisition mode x25; (b) The comparison between measured and modeled in-plane MTF for filter schemes 3 and 4.

four lines on either side of $f_y=0$ cycles/mm were averaged to compute the results. Figure 9(a) shows the results for x25 mode and filter schemes 3 and 4. The parabolic shape of the NPS is due to the reconstruction filters. The low-frequency drop is due to the RA filter, and the high-frequency drop is due to the combined effects of the SA and ST filters, both of which degrade high-frequency response.¹¹ The modeled result is in excellent agreement with the two measurements, which represent the average and the top slice. This confirms the validity of our model and shows that the assumption of shift invariance is reasonable.

Shown in Fig. 9(b) are the results for x25bin mode and filter schemes 2–4. The NPS level is the highest for scheme 2, and the measured NPS for all filter schemes agree well with the model. Due to pixel binning, the high-frequency noise is much lower than that in full resolution mode.

IV.B. In-plane MTF

Shown in Fig. 10(a) is the measured in-plane presampling MTF for all four reconstruction filter schemes. Since the zero spatial frequency response after reconstruction is not always the highest, the in-plane MTF for each reconstruction

scheme is normalized by its own maximum response. The decrease in MTF at low spatial frequencies for filter schemes 2–4 is due to the RA filter and the incomplete angular sampling in tomosynthesis. This phenomenon is not observed in CT or CBCT with complete angular sampling.⁸ The peak of the MTF curves shifts towards lower frequencies from schemes 2–4 due to the application of the SA and ST filters that reduce high-frequency response. Shown in Fig. 10(b) are the comparisons between measured and modeled in-plane MTF for filter schemes 3–4, which are most frequently used in our study for noise reduction purposes. Reasonable agreement between model and measurement was observed for both filter schemes. Compared with modeled results, the measured MTF is lower at higher frequencies. The discrepancy is more pronounced in scheme 3 when more high-frequency component is allowed through. Here, we will provide possible mechanisms that could qualitatively explain the discrepancy: (1) The modeled in-plane MTF was calculated by integrating the 3D MTF in the z -direction, which is valid for an edge phantom with infinitesimal thickness. The finite thickness (0.2 mm) of the edge phantom in our measurement leads to reduced high-frequency components in the z -direction and hence a lower in-plane MTF. However, with the application of the ST filter, this factor should be negligible since the thickness of the edge is only 1/5 of the slice thickness. (2) The MTF measurement is phase dependent because the slice thickness is larger than the phantom thickness.^{34,35} The calculation assumes the edge phantom is in perfect alignment with the center of the slice. In practice, this is difficult to achieve and there is usually a slight shift, which results in a measured in-plane MTF that is position dependent and could be lower than the modeled result. Since the ST filter removes aliasing in the z -direction, it makes the in-plane MTF phase independent. Therefore, the agreement between measured and modeled results is better with filter scheme 4. One of our ongoing investigations is to measure the presampling 3D MTF of a DBT without the effects of phase dependence and thickness of phantom. We hope to resolve the discrepancy between model and measurement in the near future.

Shown in Fig. 11 are the comparisons of the measured in-plane presampling MTF with filter schemes 3 and 4 for two different angular ranges of $\pm 10^\circ$ and $\pm 20^\circ$. The projection data from the same tomosynthesis scan were used to reconstruct the in-plane images for both cases, with the 13 central views selected for $\pm 10^\circ$. It shows that an increased angular range improves the MTF at low frequencies. This is because the in-plane MTF corresponds to the integration of the 3D MTF in the z -direction; as a result the slope of the MTF at low frequencies is proportional to the angular range of tomosynthesis sampling. This aspect is consistent with the model calculation and simulation in our companion paper, which also provides a more detailed explanation of the mechanism.²⁶ Further increase of the angular range in DBT is difficult due to the limitation in hardware design: (1) a larger fraction of the x-ray projection will fall outside of the stationary detector; (2) larger focal spot blur if the number of views and scan time are kept the same.

FIG. 11. Comparison between measured in-plane presampling with two different angular ranges, $\pm 10^\circ$ and $\pm 20^\circ$, for filter schemes 3 and 4.

IV.C. Phantom study

Shown in Fig. 12 is a projection image of the ACR phantom. The circles highlight the location of the mass and calcification features used in our quantitative analysis (cal-1, mass-1, and mass-2).

In this section, we will first present the dependence of ACR phantom images on reconstruction filters, and then on DBT acquisition parameters, i.e., angular range and detector operational modes.

IV.C.1. Dependence on reconstruction filters

Shown in Fig. 13(a) are the comparisons of the in-plane images of the ACR features using four different filter schemes with x25bin mode. It shows that the image obtained

FIG. 12. ACR phantom image obtained using the screening mode of the prototype DBT system with 28 kVp W/Rh and 140 mAs.

(a)

(b)

FIG. 13. (a) ACR phantom image reconstructed with the four schemes. The acquisition mode was x25bin with total 144 mAs. The scan direction is along the horizontal direction of the image. (b) Normalized intensity profile for cal-1 with filter schemes 1 and 4. The acquisition mode was x25bin with a total exposure of 144 mAs.

using filter scheme 1 is blurry, leading to difficulty in the detection of the third group of calcification specks. Using the other filter schemes (2–4), all the features that are visible in screening mammography (Fig. 12) can be seen in the reconstructed in-plane image. The reconstructed images of the mass with filter schemes 2–4 show a clear edge enhancement along the tube travel direction (horizontal direction in the image). This is due to the application of the RA filter to the image data acquired with a limited angular range, and has been seen quantitatively in the modeled and measured in-plane MTF (Fig. 10). Compared with schemes 3 and 4, the noise level is higher in scheme 2, which is expected from its lack of noise reduction at high frequencies.

IV.C.1.a. Calcifications. Plotted in Fig. 13(b) are the normalized profiles of cal-1 for filter schemes 1 and 4. Edge

TABLE IV. SDNR and background noise for mass-1 and mass-2 for the four reconstruction schemes acquired in x25bin with total 144 mAs.

SDNR	Filter scheme 1	Filter scheme 2	Filter scheme 3	Filter scheme 4
Mass-1	9.921	0.995	1.615	2.535
Mass-2	6.390	0.727	1.260	1.970
σ_{bak}	21.28	93.49	70.58	58.54

enhancement (shown as undershoot and overshoot) is observed for scheme 4, but not for SBP (scheme 1). The profiles for filter schemes 2 and 3 exhibit the same edge enhancement as for scheme 4, and therefore are not shown in the figure for clarity. The difference between schemes 2–4 is in their noise levels. The standard deviation σ_{bak_N} calculated using the normalized background in Fig. 13(b) is 0.071, 0.042, and 0.030 for scheme 2, 3, and 4, respectively. This is to be compared with the $\sigma_{\text{bak}_N}=0.041$ for scheme 1.

IV.C.1.b. Mass. The measured SDNR for the two mass features with different reconstruction filters is listed in Table IV. Since SDNR changes as a function of both signal difference and noise, the background noise level σ_{bak} [Eq. (5)] is also listed in Table IV to facilitate comparison. The SDNR for mass-1 is larger than that for mass-2, as expected from the higher inherent contrast in mass-1. Scheme 1 has the highest SDNR in all reconstruction schemes due to two reasons: (1) mass contains mainly low-frequency information, where scheme 1 (SBP) has the highest response; (2) the poor high-frequency response of scheme 1 effectively reduces noise. With the application of the SA and ST filters for schemes 3 and 4, there is significant improvement in SDNR compared to scheme 2 due to the suppression of noise. The background noise σ_{bak} decreases by $\sim 50\%$ from scheme 2 to scheme 4. In addition to the SDNR, the edge enhancement of the mass features is expected to help the detection with FBP.

IV.C.2. Dependence on the angular range and detector operational mode

The reconstructed images with four different combinations of angular range and detector operational mode are shown in Fig. 14(a). All the images were acquired with comparable total exposure (144–149 mAs) and reconstructed with filter scheme 4. It should be noted that the FSB due to focal spot motion for the angular range of $\pm 10^\circ$ is twice as large as that at $\pm 20^\circ$. This is because in the former case, the radiation dose was delivered with 13 views, resulting in doubled exposure time for each view. The difference in FSB could have an impact on the detection of small calcifications.

IV.C.2.a. Calcification. The normalized pixel intensity profiles for the images of cal-1 in Fig. 14(a) are shown in Fig. 14(b). Since the plots overlap significantly, only the results for the full resolution mode with filter scheme 4 are shown for clarity. It shows that with the signal intensity of cal-1 normalized to the same amplitude, the background noise with an angular range of $\pm 20^\circ$ ($\sigma_{\text{bak}_N}=0.084$) is higher than that for $\pm 10^\circ$ ($\sigma_{\text{bak}_N}=0.052$). The present study

(a)

(b)

FIG. 14. (a) ACR phantom image reconstructed with four acquisition modes: (1) binning mode with $\pm 20^\circ$; (2) binning mode with $\pm 10^\circ$; (3) full resolution with $\pm 20^\circ$; (4) full resolution with $\pm 10^\circ$. The images are with the same angular separation between views, comparable total mAs (144–149 mAs), and were reconstructed using filter scheme 4. (b) Normalized profile for cal-1 for two angular ranges, $\pm 10^\circ$ and $\pm 20^\circ$, reconstructed using filter scheme 4. The acquisition mode was x25 with total exposure of 144 mAs.

of the detectability of calcification and its dependence on angular range and detector resolution is limited by the ACR phantom, which is designed for screening mammography. The calcification features are made from aluminum oxide (Al_2O_3), whose attenuation is lower than the actual microcalcifications in the breast. The smallest calcifications visible on screen mammography using 28 kVp W/Rh spectrum is 0.24 mm (the fourth group), which was detectable by FBP reconstruction with all detector operational modes and imaging geometry. A wider range of calcification sizes is needed for further investigation of possible advantages of certain detector operational modes and angular range in the detection of small objects.

TABLE V. SDNR and background noise for mass-1 and mass-2 with four DBT imaging modes with comparable total mAs (144–149 mAs) and filter scheme 4.

SDNR	Binning, $\pm 20^\circ$	Binning, $\pm 10^\circ$	Full resolution, $\pm 20^\circ$	Full resolution, $\pm 10^\circ$
Mass-1	2.535	1.392	0.827	0.599
Mass-2	1.970	1.349	0.683	0.496
σ_{bak}	58.54	44.53	120.05	81.31

IV.C.2.b. Mass. The calculated SDNR and background noise for mass-1 and mass-2 are shown in Table V. Doubling of the angular range to $\pm 20^\circ$ increases the SDNR by $\sim 30\%$ despite its higher background noise level. This indicates that the image quality of large, low-contrast objects such as masses can be improved by increasing the angular range, which improves the low-frequency response. This result is consistent with the MTF measurement in Fig. 11. The SDNR in binning mode is almost three times that of the full resolution mode. This is mainly due to the twofold decrease in σ_{bak} by pixel binning,²⁸ and the combined effects of binning and reconstruction filters on the signal spectrum of the objects.

V. CONCLUSION

We investigated the spatial frequency-domain image quality metrics for a prototype DBT system. The 3D NPS and in-plane presampling MTF were investigated as a function of acquisition geometry, detector operational mode, and reconstruction filters. The measured NPS and MTF have reasonable agreement with theoretical calculation using a cascaded linear system model for DBT. The image characteristics of the objects (mass and calcifications) on an ACR phantom show a good correlation with the quantitative measurement of NPS and MTF. With the FBP reconstruction method, the detection of masses can be improved by binning and increasing the angular range. Improvement in mass detection can also be achieved by the application of the SA and ST filters, which reduces high-frequency noise and eliminates aliasing.

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¹ L. T. Niklason *et al.*, "Digital tomosynthesis in breast imaging," *Radiology* **205**, 399–406 (1997).

² T. Wu, R. H. Moore, A. B. Elizabeth, A. Rafferty, and D. B. Kopans, "Breast tomosynthesis: Methods and applications," in *Categorical Courses in Diagnostic Radiology Physics: Advances in Breast Imaging: Physics, Technology, and Clinical Applications*, edited by A. Karellas and M. L. Giger (RSNA Publication, Oak Brook, IL, 2004), pp. 149–165.

³ B. Ren *et al.*, "Design and performance of the prototype full field breast tomosynthesis system with selenium based flat panel detector," *Proc. SPIE* **5745**, 550–561 (2005).

⁴ M. Bissonnette *et al.*, "Digital breast tomosynthesis using an amorphous selenium flat panel detector," *Proc. SPIE* **5745**, 529–540 (2005).

⁵ B. Zhao and W. Zhao, "Characterization of a direct full-field flat-panel digital mammography detector," *Proc. SPIE* **5030**, 157–167 (2003).

⁶ W. Zhao and J. A. Rowlands, "Digital radiology using active matrix read-out of amorphous selenium: Theoretical analysis of detective quantum efficiency," *Med. Phys.* **24**, 1819–1833 (1997).

⁷ C. J. Bischof and J. C. Ehrhardt, "Modulation transfer function of the EMI CT head scanner," *Med. Phys.* **4**, 163–167 (1977).

⁸ J. M. Boone, "Determination of the presampled MTF in computed tomography," *Med. Phys.* **28**, 356–360 (2001).

⁹ K. Yang, A. L. C. Kwan, and J. M. Boone, "Computer modeling of the spatial resolution properties of a dedicated breast CT system," *Med. Phys.* **34**, 2059–2069 (2007).

¹⁰ M. Thornton and Michael J. Flynn, "Measurement of the spatial resolution of a clinical volumetric computed tomography scanner using a sphere phantom," *Proc. SPIE* **6142**, 61421Z (2006).

¹¹ M. F. Kijewski and A. Judy, "The noise power spectrum of CT images," *Phys. Med. Biol.* **32**, 565 (1987).

¹² J. H. Siewerdsen and D. A. Jaffray, "Three-dimensional NEQ transfer characteristics of volume CT using direct- and indirect-detection flat-panel imagers," *Proc. SPIE* **5030**, 92–102 (2003).

¹³ Y. Chen, J. Y. Lo, and J. T. Dobbins III, "Noise power spectrum analysis for several digital breast tomosynthesis reconstruction algorithms," *Proc. SPIE* **6142**, 614259–614258 (2006).

¹⁴ Y. Chen, J. Y. Lo, and J. T. Dobbins III, "Impulse response analysis for several digital tomosynthesis mammography reconstruction algorithms," *Proc. SPIE* **5745**, 541–549 (2005).

¹⁵ G. Avinash, K. Israni, and B. Li, "Characterization of point spread function in linear digital tomosynthesis: A simulation study," *Proc. SPIE* **6142**, 614258–614258 (2006).

¹⁶ J. Zhou, B. Zhao, and W. Zhao, "A computer simulation platform for the optimization of a breast tomosynthesis system," *Med. Phys.* **34**, 1098–1109 (2007).

¹⁷ A. Maidment *et al.*, "Evaluation of a photon-counting breast tomosynthesis imaging system," *Proc. SPIE* **5745**, 572–582 (2005).

¹⁸ A. D. A. Maidment *et al.*, "Evaluation of a photon-counting breast tomosynthesis imaging system," *Proc. SPIE* **6142**, 61420B–61411 (2006).

¹⁹ B. Li *et al.*, "Optimization of slice sensitivity profile for radiographic tomosynthesis," *Med. Phys.* **34**, 2907–2916 (2007).

²⁰ M. J. Flynn, R. McGee, and J. Blechinger, "Spatial resolution of x-ray tomosynthesis in relation to computed tomography for coronal/sagittal images of the knee," *Proc. SPIE* **6510**, 65100D–65109 (2007).

²¹ T. Wu *et al.*, "A comparison of reconstruction algorithms for breast tomosynthesis," *Med. Phys.* **31**, 2636–2647 (2004).

²² Y. Zhang *et al.*, "A comparative study of limited-angle cone-beam reconstruction methods for breast tomosynthesis," *Med. Phys.* **33**, 3781–3795 (2006).

²³ Y. Zhang *et al.*, "Investigation of the Z-axis resolution of breast tomosynthesis mammography systems," *Proc. SPIE* **6510**, 65104A–65108 (2007).

²⁴ T. Deller *et al.*, "Effect of acquisition parameters on image quality in digital tomosynthesis," *Proc. SPIE* **6510**, 65101L–65111 (2007).

²⁵ T. Wu *et al.*, "Tomographic mammography using a limited number of low-dose cone-beam projection images," *Med. Phys.* **30**, 365–380 (2003).

²⁶ B. Zhao and W. Zhao, "Three-dimensional linear system analysis for breast tomosynthesis," *Med. Phys.* **35**, 5219–5232 (2008).

²⁷ A. C. Kak and Malcolm Slaney, *Principles of Computerized Tomographic Imaging* (IEEE Press, New York, 1988).

²⁸ W. Zhao *et al.*, "Optimization of detector operation and imaging geometry for breast tomosynthesis," *Proc. SPIE* **6510**, 65101M–65112 (2007).

²⁹ B. Zhao and W. Zhao, "Imaging performance of an amorphous selenium digital mammography detector in a breast tomosynthesis system," *Med. Phys.* **35**, 1978–1987 (2008).

³⁰ T. Mertelmeier, J. Orman, W. Haerer, and M. K. Dudam, "Optimizing filtered backprojection reconstruction for a breast tomosynthesis prototype device," *Proc. SPIE* **6142**, 61420F (2006).

³¹ E. Samei, M. J. Flynn, and D. A. Reimann, "A method for measuring the presampled MTF of digital radiographic systems using an edge test device," *Med. Phys.* **25**, 102–113 (1998).

³² A.-K. Carton *et al.*, "Validation of MTF measurement for digital mammography quality control," *Med. Phys.* **32**, 1684–1695 (2005).

³³ A. Maidment and M. Albert, "Conditioning data for calculation of the

- modulation transfer function," *Med. Phys.* **30**, 248–253 (2003).
- ³⁴J. T. Dobbins III, "Effects of undersampling on the proper interpretation of modulation transfer function, noise power spectra, and noise equivalent quanta of digital imaging systems," *Med. Phys.* **22**, 171–181 (1995).
- ³⁵J. A. Segui and W. Zhao, "Amorphous selenium flat panel detectors for digital mammography: Validation of a NPWE model observer with CD-MAM observer performance experiments," *Med. Phys.* **33**, 3711–3722 (2006).