Effects on image quality of a 2D antiscatter grid in x-ray digital breast tomosynthesis: Initial experience using the dual modality (x-ray and molecular) breast tomosynthesis scanner

Tushita Patel^{a)}

Department of Physics, University of Virginia, Charlottesville, Virginia 22904

Heather Peppard

Department of Radiology and Medical Imaging, University of Virginia, Charlottesville, Virginia 22908

Mark B. Williams

Department of Physics, University of Virginia, Charlottesville, Virginia 22904; Department of Radiology and Medical Imaging, University of Virginia, Charlottesville, Virginia 22908; and Department of Biomedical Engineering, University of Virginia, Charlottesville, Virginia 22908

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Purpose: Radiation scattered from the breast in digital breast tomosynthesis (DBT) causes image degradation, including loss of contrast between cancerous and background tissue. Unlike in 2-dimensional (2D) mammography, an antiscatter grid cannot readily be used in DBT because changing alignment between the tube and detector during the scan would result in unacceptable loss of primary radiation. However, in the dual modality breast tomosynthesis (DMT) scanner, which combines DBT and molecular breast tomosynthesis, the tube and detector rotate around a common axis, thereby maintaining a fixed tube-detector alignment. This C-arm geometry raises the possibility of using a 2D (cellular) focused antiscatter grid. The purpose of this study is to assess change in image quality when using an antiscatter grid in the DBT portion of a DMT scan under conditions of fixed radiation dose.

Methods: Two 2D focused prototype grids with 80 cm focal length were tested, one stack-laminated from copper (Cu) and one cast from a tungsten-polymer (W-poly). They were reciprocated using a motion scheme designed to maximize transmission of primary x-ray photons. Grid-in and grid-out scatter-to-primary ratios (SPRs) were measured for rectangular blocks of material simulating 30%, 50%, and 70% glandular tissue compositions. For assessment of changes in image quality through the addition of a grid, the Computerized Imaging Reference Systems, Inc., phantom Model 011A containing a set of 1 cm thick blocks simulating a range of glandular/adipose ratios from 0/100 to 100/0 was used. To simulate 6.5 and 8.5 cm thick compressed breasts, 1 cm thick slices of PMMA were added to the Model 011A phantom. DBT images were obtained with and without the grid, with exposure parameters fixed for a given compressed thickness. Signal-difference-to-noise ratios (SDNRs), contrast, and voxel value-based attenuation coefficients (μ) were measured for all blocks from reconstructed phantom images.

Results: For 4, 6, and 8 cm tissue-equivalent block phantom thicknesses, the inclusion of the W-poly grid reduced the SPR by factors of 5, 6, and 5.8, respectively. For the same thicknesses, the copper grid reduced the SPR by factors of 3.9, 4.5, and 4.9. For the 011A phantom, the W-poly grid raised the SDNR of the 70/30 block from 0.8, -0.32 , and -0.72 to 0.9, 0.76, and 0.062 for the 4.5, 6.5, and 8.5 cm phantoms, respectively. It raised the SDNR of the 100/0 block from 3.78, 1.95, and 1.0 to 3.79, 3.67, and 3.25 for the 4.5, 6.5, and 8.5 cm phantoms, respectively. Inclusion of the W-poly grid improved the accuracy of image-based μ values for all block compositions. However, smearing of attenuation across slices due to limited angular sampling decreases the sensitivity of voxel values to changing composition compared to theoretical μ values.

Conclusions: Under conditions of fixed radiation dose to the breast, use of a 2D focused grid increased contrast, SDNR, and accuracy of estimated attenuation for mass-simulating block compositions in all phantom thicknesses tested, with the degree of improvement depending upon material composition. A 2D antiscatter grid can be usefully incorporated in DBT systems that employ fully isocentric tube-detector rotation. © 2016 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4943632>]

Key words: antiscatter grid, scatter, digital breast tomosynthesis, SDNR, contrast

1. INTRODUCTION

Medical x-ray imaging provides structural information by exploiting the differing x-ray attenuation properties of various tissue types with the goal of differentiating between abnormal and healthy tissues. However, the radiographic attenuation of the tissue in malignant breast masses is very similar to that of normal fibroglandular breast tissue, making sensitive distinction of small differences in attenuation crucial in all x-ray based breast imaging.

In digital breast tomosynthesis (DBT) imaging, the x-ray tube is moved relative to the breast in order to obtain a series of low-dose projection images over a limited angular range from which a 3D image of the breast can be reconstructed. In conjunction with two-dimensional (2D) full field digital mammography (FFDM), DBT has been shown to reduce recall rates and improve positive predictive value and cancer detection rate in a screening population, especially for women with radiodense breasts.^{1,2} Like x-ray computed tomography (CT), DBT images consist of voxels whose values are ideally linearly related to the linear attenuation coefficient μ of the tissue within the voxel. However, the limited angular range of DBT does not provide a complete tomographic data set. Therefore, the estimate of μ for a given voxel is affected by artifactual smearing of attenuation from voxels lying in nearby slices.^{3,4}

Scattered radiation in projection x-ray imaging decreases lesion contrast and increases image noise. In tomographic x-ray imaging, such as CT, it produces cupping and streak artifacts and causes errors in the measurement of the spatially varying x-ray attenuation coefficients (i.e., errors in the voxel values). Scatter increases strongly with increasing object thickness.⁵ Scatter in DBT introduces additional errors in the form of artifactual spatially varying biases in voxel value and attenuation quantitation errors, similarly to those in CT.⁶⁻⁹ Using Monte Carlo simulations, Wu *et al.* found that for a 5 cm thick breast image contrast for a 1.4 cm mass is reduced by 30%, the voxel value is reduced by 28%, and the signal-difference-to-noise ratio (SDNR) is reduced by 60%.¹⁰

To minimize the negative effects of scatter in 2D FFDM, antiscatter grids are routinely placed between the breast and the detector to block a portion of the scatter that would otherwise reach the detector. One dimensional (1D) grids employ parallel strips of highly attenuating material, such as lead, separated by low-attenuation material such as carbon fiber. To minimize cutoff of primary (nonscattered) rays by the grid walls, most 1D antiscatter grids are focused, wherein the strip orientations are increasingly non-normal with increasing distance from the grid midline so that the midlines of the interstrip apertures intersect at a line that, in turn, intersects the x-ray focal spot (XFS). For improved scatter rejection, 2D focused grids are used¹¹ whose holes are oriented such that their central axes all intersect at a single point in space, called the grid focal point (GFP). The GFP is located a fixed distance (the focal length) measured perpendicularly from the midpoint of the grid's posterior (chest wall) edge. When a focused 2D grid is well-aligned, all holes of the grid point

toward the XFS, and the GFP and XFS are located at the same point in space.

While both 1D and 2D antiscatter grids are available for FFDM, these grids cannot readily be used in most clinical tomosynthesis systems. During DBT scans, the x-ray detector typically is stationary or rotated through a small angle and thus does not maintain a fixed alignment with respect to the moving x-ray source. Therefore, either a conventional 1D grid, whose lamellae are oriented in the anterior-posterior direction, or a 2D grid, whose lamellae are oriented at an angle with respect to the detector matrix, would produce unacceptably large attenuation of primary (unscattered) radiation for most DBT tube positions.

In contrast, focused 2D antiscatter grids can be used in cone-beam computed tomography breast imaging (CTBI) systems, which are also plagued by scatter. Rather than single row CT, CTBI systems make use of multiple detector rows for wider fields of view (FOVs), further increasing the importance of scatter rejection with wider cone-beam angles.^{9,12,13} In CTBI, the amount of primary beam attenuated by the lamellae of a 2D grid does not vary with changing viewing angle because the detector and source rotate together about a single axis that runs through the pendant breast.¹⁵ 2D antiscatter grids have been shown to drastically reduce scatter from CTBI images, leading to marked improvement in image quality.¹⁶

The dual modality breast tomosynthesis (DMT) scanner developed at the University of Virginia is an investigational imaging system that combines DBT and nuclear medicine emission tomosynthesis, or molecular breast tomosynthesis (MBT) on a single gantry.¹⁷ Much like CTBI, the DMT x-ray tube and detector rotate around a common axis. This C-arm geometry keeps the orientation of the detector and tube fixed and makes DBT equivalent to limited angle CT using a third generation CT gantry. This geometry raises the possibility of using a 2D (cellular) focused antiscatter grid, as has recently been employed in cone-beam or multirow CT scanners.¹⁸

Both 1D and 2D radiographic grids are reciprocated, i.e., translated in the left-right direction in a plane parallel to the detector surface, during x-ray exposure in order to minimize the appearance of the shadow of the grid itself in the image. A grid reciprocation system has been built onto the DMT scanner for acquiring tomosynthesis projection images with a focused antiscatter grid using a nonlinear motion scheme for increasing primary transmission. Details of this design are described in a separate paper (Design and evaluation of a grid reciprocation scheme for use in digital breast tomosynthesis). A prototype 2D focused grid was fabricated for this study by Mikro Systems, Inc. It was made using a stack lamination process to generate a mold from which a tungsten-polymer (W-poly) composite grid was formed. Similar W-poly 2D focused grids have been proven to be a promising tool for reducing scatter in CT.¹⁸ In that application, the grid is nonreciprocating and aligned precisely with the detector matrix such that the grid lamellae overlaid the interdetector gaps. The purpose of this study is to assess the improvement in image quality when using a reciprocating antiscatter grid in the DBT portion of a DMT scan.

2. MATERIALS AND METHODS

2.A. Image acquisition

X-ray images were taken using the high-sensitivity mode¹⁹ of the 2923MAM x-ray detector manufactured by Dexela/Perkin-Elmer (London). The x-ray source was a Varian tube with a tungsten target and a 50 μm thick rhodium filter. The source to image distance (SID) is 82 cm. The breast is compressed on a support that is located 14 cm above the x-ray detector. The axis of rotation (AOR) for both x-ray detector and x-ray source is located 4 cm above the breast support. Figure 1 is a schematic of the DMT system geometry.

Two focused 2D antiscatter grid prototypes were used for these experiments. The first was a W-poly grid with a height of 3.65 mm and the second was a copper (Cu) grid with a 3.85 mm height. Both contain square grid holes with 1 mm hole size and 0.1 mm septal thickness, have an active area of 17 $\times$ 26 cm, and have a focal length of 80 cm. Beam-stop experiments were performed with both grids for determination of scatter-to-primary ratio (SPR). All other data presented here were obtained using only the W-poly grid. The grids were positioned at the central chest wall edge of the detector, and all tests were performed using only the portion of the detector's 23 $\times$ 29 cm area that was covered by the grid. A voice coil motor, manufactured by H2W Technologies, Inc., was used for grid reciprocation.

The Computerized Imaging Reference Systems, Inc. (CIRS) phantom Model 011A (Ref. 14) was chosen for comparing grid-in DBT image quality to grid-out DBT image

quality. This model simulates a breast of 50% glandular/50% adipose (50/50 composition) surrounded by a 5 mm thick layer of fat-simulating material. Five 1 cm blocks are embedded in this phantom, simulating tissue compositions of 0% glandular/100% adipose (0/100), 30% glandular/70% adipose (30/70), 50% glandular/50% adipose (50/50), 70% glandular/30% adipose (70/30), and 100% glandular/0% adipose (100/0). A range of overall phantom thicknesses were tested by placing additional PMMA onto the 4.5 cm CIRS phantom.

For a given phantom thickness, all imaging parameters and phantom positions were kept fixed for both grid-in and grid-out images, the only difference in the two acquisitions being the presence or absence of a grid prototype. The DBT exposure parameters for this work were adjusted so that the resulting average glandular doses (AGDs) were similar to those from the clinical screening mammograms of nine DMT study subjects whose breast compositions were approximately 50% glandular and 50% adipose. These subjects were chosen because the scanners used to obtain their clinical mammograms also used a tungsten target/rhodium filter combination. Figure 2 shows the AGDs reported in the DICOM header of the screening mammograms plotted versus the compressed thickness from 16 images.

A linear fit to this data was applied, and the resulting relation provided an approximate AGD for a given value of compressed thickness as shown in Eq. (1).

$$\text{AGD} = a \cdot T_C - b. \quad (1)$$

The coefficients a and b from the linear fit of Fig. 2 have values of $a = 0.044 \text{ mGy/mm}$ and $b = 1.03 \text{ mGy}$. Average values of AGD within 1 cm width bins of compressed thickness have also been plotted in Fig. 2. The fit line was used to calculate exposure parameter settings for the phantom studies that would result in comparable AGD values for a given phantom thickness.

These total doses were divided equally among thirteen 500 ms exposures taken at $\pm 12^\circ$, $\pm 8^\circ$, $\pm 5^\circ$, $\pm 3^\circ$, $\pm 2^\circ$, $\pm 1^\circ$, and 0° relative to the direction of compression. The tube voltage was determined from a look-up table generated by application of the methods described by Williams *et al.* for maximizing the ratio of the lesion SDNR squared to the AGD

FIG. 1. Schematic of DMT DBT geometry illustrating location of the AOR of both source and detector, the SID, and the source to AOR distance.

FIG. 2. A plot of FFDM AGDs versus compressed thickness for breasts of approximately 50/50 tissue composition imaged by clinical systems with tungsten targets and rhodium filters. Individual data points are plotted as blue diamonds, with average values within 10 mm bins plotted as green triangles. The fit line is a linear least squares fit to the raw (unbinned) data.

TABLE I. Exposure parameters for CIRS Model 011A phantom.

Compressed thickness (cm)	Tube voltage (kVp)	Half-value layer (mm)	Entrance surface air kerma (mGy)	Normalized glandular dose (D_{gN}) (mGy/mGy)	Average glandular dose (mGy)	Effective beam energy (keV)
4.5	28	0.60	3.2	0.329	1.05	18.6
6.5	29	0.61	7.37	0.251	1.84	18.7
8.5	31	0.62	14.55	0.203	2.94	18.8

for a given glandular/adipose composition and compressed thickness.²⁰ Using the conversion tables of Boone²¹ and the measured half-value layers at the kVp settings determined using the method described by Williams *et al.*,²⁰ tube current and exposure times were determined to attain the target AGD for each phantom thickness. Table I shows, for 4.5, 6.5, and 8.5 cm compressed thickness and 50/50 composition, the AGD values obtained from the curve fit of Fig. 2, the kVp setting resulting from the $(\text{SDNR})^2/\text{AGD}$ maximization procedure, the normalized glandular dose (D_{gN}), and the entrance surface air kerma. Also provided in the table are the half-value layers and effective beam energies for each tube voltage setting.

Image reconstruction was performed using iterative reconstruction software provided by the manufacturer of the x-ray detector. Image reconstruction was performed using a proprietary algorithm developed by Dexela. Projection image pixel values are first converted to attenuation values by taking their negative log following normalization by dividing each by the mean pixel value corresponding to the unattenuated beam (i.e., the mean of pixels lying outside the breast region in the image). The software then uses an iterative statistical method, going through ten iterations to attempt to match line integrals through the estimated volume to the observed densities in the projections. The known compressed thickness is used as input to the algorithm for regularization with the goal of confining attenuation to the region of the breast. This is necessary given the angular undersampling inherent in tomosynthesis. The resulting images have reconstructed slice thicknesses of 1 mm and in-plane voxel dimensions of 150 μm . The output voxel values are the effective (polyenergetic) linear attenuation coefficients multiplied by a factor ~ 1000 for display purposes. All analyses of image quality were done using ImageJ.

2.B. Primary transmission measurement

The primary transmission of the grid was measured by comparing images acquired with the grid in the beam to images acquired using the same techniques but without the grid prototype in the beam. To minimize the amount of scatter in the beam, the breast support and compression paddle were not present for these measurements, and there was no other scattering or attenuating material placed in the beam. To ensure a linear detector response to input fluence and avoid detector pixel saturation, tube parameters of 28 kV and 2.5 mAs were used. The grid focal point was aligned with the x-ray focal spot by aligning the center of the chest wall edge of the stationary grid with the x-ray focal spot's projection onto the x-ray detector with the grid placed at a distance of 80 cm from the source. Five consecutive images were acquired

at a projection angle of 0° for both grid-out acquisitions and acquisitions with the reciprocating grid. Images were corrected for dark current but were not flat-field corrected.

Nine 300×300 pixel regions of interest (ROIs) were drawn over the active area of the grid, as shown in Fig. 3. To account for any changes in tube output from one exposure to the next, a factor α was calculated as described by Fetterly and Schueler²² and as shown in

$$\alpha = \left(\overline{R_{\text{Bkg, grid-out}}} / \overline{R_{\text{Bkg, grid-in}}} \right), \quad (2)$$

where $\overline{R_{\text{Bkg, grid-out}}}$ and $\overline{R_{\text{Bkg, grid-in}}}$ are the average pixel values for the grid-out and grid-in images, respectively, in a ROI lying outside the grid region. The primary transmission T_P through the grid was then calculated as

$$T_P = \alpha \overline{R_{\text{grid-in}}} / \overline{R_{\text{grid-out}}}, \quad (3)$$

where $\overline{R_{\text{grid-in}}}$ and $\overline{R_{\text{grid-out}}}$ are the mean pixel values in the ROIs of the grid-in and grid-out images, respectively, within the region of the active area of the grid.

2.C. Scatter-to-primary ratios

For characterizing the amount of scatter present in DMT DBT images, scatter-to-primary ratios were measured using

FIG. 3. Locations of ROIs for calculating primary transmission at various positions within the active area of the moving grid (solid lines) as well as the background ROI (dashed line) for calculating α in Eq. (2). The anterior side of the detector is on the left and the posterior edge is on the right side of the image.

the beam-stop method described by Fetterly and Schueler.²² Projection images were acquired of circular lead blockers placed over uniform phantoms consisting of CIRS tissue-equivalent blocks with dimensions of 10×12.5 cm and thicknesses ranging from 5 to 2 cm.³⁰ There were three sets of these blocks that simulated breast glandular/adipose compositions of 30/70, 50/50, and 70/30. For each composition, three phantoms were built from two 5 mm thick skin-simulating blocks, included at the entrance and exit surfaces, and different combinations of block thicknesses of the desired composition to obtain total phantom thicknesses of 4, 5.5, and 7 cm. The three compositions and three thicknesses generated a total of nine different phantoms. Images were first taken without any form of scatter rejection and then repeated with the grid prototypes in the beam just above the detector surface. To get an estimate of SPR for a 50/50 composition breast that would cover more of the detector's 23×29 cm FOV, the same experiments were performed with blocks of PMMA, four with dimensions of 16×24×1 cm thick and one with dimensions of 30.5×25.5×4 cm thick. There was no skin-simulating material included for the PMMA phantoms. Three PMMA phantoms were created with total compressed thicknesses of 4, 6, and 8 cm. Data from these images were then compared to projection images using the same set of CIRS and PMMA block thicknesses, but without any lead blockers, to obtain the final SPRs for both grid-in and grid-out cases.

2.D. Image quality comparison

To assess the impact of the grid on DBT images of objects whose attenuation is slightly higher than that of the surrounding tissue, as is the case for breast masses, the signal difference (SD), the SDNR, and the contrast in the reconstructed image slices were calculated for all blocks. All measurements of SD, SDNR, and contrast were made from a single slice in the reconstructed image that corresponded to the approximate center of the 1 cm blocks. Background and signal block mean voxel values were determined by Eqs. (4) and (5), respectively,

$$V_{\text{Bkg}} = \frac{1}{N} \sum_{i=0}^N v_{\text{Bkg},i}, \quad (4)$$

$$V_S = \frac{1}{N} \sum_{i=0}^N v_{S,i}, \quad (5)$$

where $v_{\text{Bkg},i}$ are the average voxel values within a 30×30 voxel² ROI drawn immediately adjacent to a given signal block, and $v_{S,i}$ are the voxel values within ROIs of the same size centered on the signal blocks. Background ROIs were drawn immediately above the signal blocks following the experiments of Liu and Li using a similar phantom.²³ The mean voxel values of Eqs. (4) and (5) were used to calculate SD, SDNR, and contrast C as defined in Eq. (6)–(8),

$$\text{SD} = V_S - V_B, \quad (6)$$

$$\text{SDNR} = \frac{\text{SD}}{\sigma_B}, \quad (7)$$

$$C = \frac{\text{SD}}{V_B}. \quad (8)$$

Noise σ_B was estimated by Eq. (9) using the standard deviation of the voxel values $v_{B,i}$ within the background ROI,

$$\sigma_B = \sqrt{\frac{1}{(N-1)} \sum_{i=1}^N (v_{B,i} - V_B)^2}. \quad (9)$$

The SD, SDNR, and contrast were measured in both grid-in and grid-out images, and their differences (grid-in minus grid-out) were calculated for each. Data from five trials of the grid-in acquisitions were averaged together for the final values.

Because the CIRS phantom has a uniform background throughout most of its volume, the reconstructed image slices should ideally have a uniform background voxel value within a given slice. However, as has been shown by Sechopoulos *et al.* and by Liu and Li, SPR increases near the periphery of the breast. This is due to the combination of (a) scatter originating in the regions of the breast support and detector cover that are exposed to the unattenuated beam, some of which is scattered under the shadow of the breast²⁴ and (b) decreased attenuation of primary photons by the thinner breast periphery.²³ To assess the variation in voxel value resulting from the spatially varying scatter, intensity profiles along the posterior–anterior direction were drawn through the 50/50 block in slices centered on the block in each phantom image. Twenty profiles were averaged together to create a single, smoother profile. Profiles were also drawn within the same slice along a direction perpendicular to the posterior-to-anterior direction. The profiles were sufficiently long to include the fat-simulating outer layer of the phantom.

2.E. Accuracy of estimated linear attenuation coefficients

To assess the accuracy of the image-based estimates of the linear attenuation coefficients μ of materials within the phantoms, voxel values within the reconstructed images were converted to estimates of μ and compared to theoretical μ values. Theoretical values μ_T of the blocks embedded in the phantom at selected energies were provided by CIRS. To calculate the approximate theoretical μ_T values for the polyenergetic spectra used for these phantom experiments, effective monoenergetic beam energies were calculated from half-value layer measurements for each kVp setting used. The measured HVLs and effective energies are listed in Table I. The effective beam energies were used to interpolate the μ versus energy data from CIRS to obtain theoretical μ_T values for each of the block compositions, which are listed in Table II.

In order to obtain DBT images of the various tissue-simulating materials under low-scatter conditions, DBT

TABLE II. Linear attenuation coefficients μ_T (cm⁻¹).

Tube voltage (kV)	0/100	30/70	50/50	70/30	100/0
28	0.67	0.77	0.84	0.80	0.91
29	0.66	0.76	0.83	0.79	0.90
31	0.65	0.75	0.81	0.78	0.88

FIG. 4. Scatter-to-primary ratios of the DMT DBT system in 0° projection images. (a) shows SPRs for uniform CIRS phantom blocks simulating 30/70, 50/50, and 70/30 breast compositions. Grid-out SPRs are represented by dashed lines and Cu grid-in SPRs are plotted as solid lines. SPRs are plotted for varying thicknesses of PMMA in (b). The dotted line shows the SPR when there is no form of scatter rejection. The solid lines plot the SPR with the W-poly (circle symbols) and Cu (triangle symbols) grids located just above the detector surface.

images were acquired of 5 mm thick blocks simulating compositions of 0/100, 30/70, 50/50, 70/30, and 100/0 compositions. These were part of the same CIRS phantom set used for the SPR experiments. In order to maintain consistency in phantom entrance beam quality for the thin (5 mm) and thicker phantom tests, the compression paddle was present during all acquisitions. However for the low-scatter (5 mm phantom) tests, the paddle was placed at a height of 16 cm above the breast support in order to minimize the amount of paddle-induced scatter reaching the detector. Low-scatter DBT images were obtained for each composition at tube voltages of 28, 29, and 31 kVp, both with and without the W-poly grid.

A linear relationship between the voxel value of the 5 mm blocks $V_{S,5mm}$ and the theoretical linear attenuation coefficient μ_T is defined as follows:

$$V_{S,5mm} = m \cdot \mu_T + k. \quad (10)$$

Thus linear fits to plots of $V_{S,5mm}$ versus μ_T permit the coefficients k and m to be found for a given kVp setting. These coefficients can then be used to calculate the image-based linear attenuation coefficients μ_S of the five blocks embedded in the CIRS 011A breast phantom from their respective mean voxel values V_S using

$$\mu_S = \frac{V_S - k}{m}. \quad (11)$$

3. RESULTS

3.A. Primary transmission and scatter-to-primary ratios

The highest primary transmission was through the 300×300 pixel region beneath the focal spot, which was found to be 75.6% for the W-poly grid and 73.5% for the copper grid. Minimum values of 70% were observed for both the W-poly and copper grids within the ROIs drawn at the center of the anterior side of the image. When averaging over all

ROIs that covered the grid's active area, the average primary transmission was found to be 72% for both prototypes.

Figure 4(a) shows SPRs in the 0° projection view plotted versus phantom thickness for the 10×12.5 cm CIRS phantom blocks. The dashed lines with cross (x) symbols, circles, and diamonds represent SPRs for 30/70, 50/50, and 70/30 breast compositions, respectively, when there is no form of scatter rejection. Corresponding data taken with the Cu grid prototype are represented by solid lines connected by triangle symbols, asterisks, and square symbols for the 30/70, 50/50, and 70/30 compositions, respectively. SPR data for the larger PMMA phantoms with thicknesses of 4, 6, and 8 cm are shown in Fig. 4(b). The dashed line with the triangles is a plot of the SPRs when there is no form of scatter rejection. The solid lines with the circle and triangle symbols show the SPR when the W-poly grid and Cu grid prototypes, respectively, were present in the x-ray beam. Because the W-poly grid provided better scatter rejection for approximately the same primary transmission as the Cu grid, all grid-in results presented in the remainder of the paper were obtained with the W-poly grid.

3.B. Reconstructed slices

Figure 5 shows the reconstructed slices from which all SD, SDNR, and contrast data were measured. Images where a grid was not present during acquisition are shown in the left column of Figs. 5(a), 5(c), and 5(e) for the 4.5, 6.5, and 8.5 cm thick phantoms. The corresponding slices from DBT acquisitions of the same phantoms taken with the W-poly grid in the beam are shown in the right column of Figs. 5(b), 5(d), and 5(f) for 4.5, 6.5, and 8.5 cm. The greater uniformity in background voxel value with the grid present is especially evident for the 6.5 and 8.5 cm phantoms. Fibers located below the 1 cm blocks and masses in the 8.5 cm phantom are more visible in the grid-in image than without the grid.

Figure 6 shows regions of slices in which some of the speck groups and spherical masses embedded in the phantom are most in focus. Grid-out slices are shown in Figs. 6(a), 6(c),

FIG. 5. Reconstructed slices of the phantoms built from the CIRS Model 011A phantom and additional scattering material. Slices from acquisitions without the grid prototype are shown in (a), (c), and (e) for the 4.5, 6.5, and 8.5 cm phantoms, respectively. The corresponding slices from the acquisitions with the W-poly grid are shown in (b), (d), and (f) for the 4.5, 6.5, and 8.5 cm phantoms, respectively. The posterior edge of the detector is at the top of each image.

and 6(e) for the 4.5, 6.5, and 8.5 cm phantoms, respectively, and grid-in slices are shown in Figs. 6(b), 6(d), and 6(f) for 4.5, 6.5, and 8.5 cm. The window and level settings for each of the images in Fig. 6 were set to maximize the visibility of the speck groups. As a result, the 1 cm blocks of varying glandularity and the smaller masses at the bottom of the grid-out images become less visible with increasing phantom thickness.

Figures 5 and 6 illustrate the increasing nonuniformity of background voxel values with increasing phantom thickness, making it difficult to see the masses in the 8.5 cm phantom when the display settings are adjusted as in Fig. 6(e). Voxel values in the left-right direction are nearly constant over

the smaller region where the signal blocks are located. Inclusion of the antiscatter grid results in more uniform voxel values in the anterior-posterior direction, permitting improved visualization of speck groups, fibers, and masses that are closer to the anterior side of the phantom as shown in Figs. 5 and 6. For example, at least four fibers can be seen in all of the grid-in images in Fig. 5, whereas only two to three fibers can be identified in the 6.5 and 8.5 cm phantoms without the grid. The nonuniformity of voxel values in the grid-out images of Figs. 6(c) and 6(e) causes the box that contains the fibers to be nearly undetectable at the selected window/level settings. Moreover, in Fig. 6, it is easier to find the two speck groups above the white circle, the signal blocks, and the two smaller

FIG. 6. Reconstructed slices showing a zoomed view of some of the speck groups and spherical masses embedded in the phantom. Grid-out slices are shown in (a), (c), and (e) for the 4.5, 6.5, and 8.5 cm phantoms, respectively. The corresponding W-poly grid-in slices are shown in (b), (d), and (f) for the 4.5, 6.5, and 8.5 cm phantoms, respectively. The posterior edge is at the top of each image.

masses to the left of the white circle in the grid-in images of the 6.5 and 8.5 cm phantoms than in the corresponding grid-out images.

3.C. Line profiles through uniform background

Figure 7(a) is a region of a reconstructed slice of the 4.5 cm grid-out image with a red region indicating adjacent columns

of voxels that were averaged to create a line profile through the 50/50 block. The anterior side of the phantom is at the bottom of the image. The plots in Figs. 7(b)–7(d) are the averaged profiles through grid-out and grid-in images of the 4.5, 6.5, and 8.5 cm phantoms, respectively. The zero voxel position is at the posterior end of the red region in Fig. 7(a), and the 670 position is at the anterior end of the 50/50 phantom background region (it does not include the fat layer).

FIG. 7. (a) is a reconstructed slice of the 4.5 cm phantom showing a region (red) used to create an average profile through the center of the phantom. The anterior side of the phantom is at the bottom of the image. (b)–(d) show the profiles for the 4.5, 6.5, and 8.5 cm phantoms, respectively. Grid-out data are illustrated by the dashed line and grid-in data are shown by the solid line. Voxel position 0 is the posterior-most end of each profile. Spikes in the profiles correspond to the circumference of the white circle located in the phantom above the signal blocks and the depression in values for voxels ~400–475 values is caused by the lower attenuation square region containing nylon fibers (see color online version).

Similarly, Fig. 8(a) shows the same slice as in Fig. 7(a), but containing a red region used to create an averaged profile in the left-to-right direction parallel to the posterior edge of the detector. Figures 8(b)–8(d) are the averaged profiles through the 4.5, 6.5, and 8.5 cm phantoms, respectively. The fat layer of the phantom begins at approximately voxel position 20 and transitions to the uniform 50/50 phantom background near voxel position 70 where a shoulder in the profile can be seen.

3.D. SD, SDNR, and contrast

Figure 9 contains plots of the SD, SDNR, and contrast for all signal blocks in all phantoms tested. Figure 9(a) is an example image showing where the ROIs were drawn over the signal blocks (solid lines) and in the background regions above the signal blocks (dotted lines). The rightmost signal block is drawn over the 100/0 block. Since the phantoms were not altered or moved between grid-in and grid-out acquisitions, the ROI

positions and sizes are identical for a given tested thickness. Signal block ROIs were drawn small enough to avoid including bright or dark edge artifacts occurring at the sharp left-to-right interfaces between blocks [see Fig. 9(a)]. Measured SD, SDNR, and contrast values for each block type are plotted versus glandularity in Figs. 9(b)–9(d), respectively, for 4.5, 6.5, and 8.5 cm phantoms. Differences in SD, SDNR, and contrast (grid-in minus grid-out) were calculated for each block and are plotted in Figs. 10(a)–10(c), respectively, for all phantoms.

To illustrate the change in the image noise resulting from use of the grid, Fig. 11 is a plot of the standard deviation σ_B of the voxel values within the 50/50 blocks plotted versus compressed thickness for all phantoms tested. Values of σ_B for the grid-in acquisitions are plotted as circles connected by a solid line and those for the grid-out acquisitions by triangles connected by a dashed line. The plot shows that the RMS noise in the grid-in image slices is 20%, 26%, and 22% higher than that in the grid-out images for phantom thicknesses of 4.5, 6.5, and 8.5 cm, respectively.

Fig. 8. (a) is a reconstructed slice of the 4.5 cm phantom showing the red region in which columns were averaged to create a profile through the uniform background in the direction parallel to the posterior edge of the detector. The window/level settings have been chosen to permit visualization of the 0/100 composition skin layer at the phantom periphery. (b)–(d) show the profiles for the 4.5, 6.5, and 8.5 cm plots, respectively. In each case, the grid-in profile has a larger average voxel value. The left side of the image corresponds to the 0 position on the profile plots (see color version online).

3.E. Estimated attenuation coefficient

Figure 12 shows the mean voxel values $V_{S,5mm}$ of the 5 mm blocks simulating 0/100, 30/70, and 50/50, 70/30, and 100/0 compositions, plotted versus their theoretical linear attenuation coefficients μ_T . Figures 12(a)–12(c) are for tube voltages of 28, 29, and 31 kVp, respectively. In each graph, the $V_{S,5mm}$ values measured both with and without the W-poly grid are plotted. These plots were then used to generate estimates of the linear attenuation coefficients of the 1 cm blocks embedded in the CIRS 011A breast phantom using Eqs. (10) and (11). A linear fit was applied to each plot of Fig. 12 to determine, for each kVp setting, the constants k and m in Eq. (10). Fitted values of constants k and m are given in Table III for each kVp, as measured either with or without the antiscatter grid. The grid-in and grid-out coefficient values differ slightly because a small but non-negligible amount of scatter was generated by the 5 mm thick phantom blocks. Grid-in voxel-based μ_S were determined from the grid-in linear fit parameters, and grid-out voxel-based μ_S were determined from the grid-out linear fit parameters.

Figures 13(a)–13(c) are plots of the theoretical linear attenuation coefficients along with the voxel-based μ_S values for the 1 cm blocks embedded in the CIRS 011A phantom derived using the constants in Table II and Eq. (11). Plots are shown for 4.5, 6.5, and 8.5 cm overall phantom thickness, respectively, for both grid-in and grid-out conditions. For the thicker (6.5 and 8.5 cm) phantoms, the removal of scatter by the grid elevates the μ_S values for all block compositions. However, comparison to the theoretically expected values shows that the μ_S values of blocks with glandularities lower than those of the surrounding 50/50 composition material of the 011A phantom are higher than expected, whereas those of lower glandularity blocks are lower than expected. This phenomenon is a result of artifactual smearing of background attenuation information into the block slices during DBT reconstruction. The artifacts are a result of the limited acquisition angular range (24°) used here. Figure 14 shows the resulting percent errors relative to the theoretical values, in the image-based linear attenuation coefficients for the 4.5, 6.5, and 8.5 cm phantoms, respectively.

FIG. 9. (a) shows the locations of the ROIs drawn for measurements of SD, SDNR, and contrast. The dashed squares in (a) show the ROIs used for determining the mean background voxel value and background noise for each adjacent signal block, and the solid ROIs, from left to right, show the 0/100, 30/70, 50/50, 70/30, and 100/0 adipose/fibroglandular signal block ROIs. (b) SD, (c) SDNR, and (d) contrast of all embedded 1 cm signal blocks versus % glandularity of each block for the 4.5, 6.5, and 8.5 cm phantoms. Grid-out data are plotted using dashed lines and grid-in data are plotted using solid lines.

4. DISCUSSION

Despite the presence of an air gap of ~ 15 cm between the detector and the breast support, Fig. 4 shows that there is still a large amount of scatter detected in DMT projections images for all phantom thicknesses and compositions evaluated. While scattered radiation dominates the image in the grid-out case for compressed thicknesses above ~ 7 cm, the grid prototype rejects the majority of the scatter. Grid-in SPRs are nearly constant (~ 0.2) over all tested thicknesses, with a magnitude approximately 17%–20% of that with no grid for PMMA phantoms of 4–8 cm thickness.

In one of the relatively few experimental studies of the use of an antiscatter grid in tomosynthesis, King *et al.* used phantoms to investigate the impact of a grid on image quality and radiation dose in a commercial system (Definium

8000, GE Healthcare) equipped with removable 1D grids and automatic exposure control (AEC). Using acrylic phantoms ranging in thickness from 10 to 25 cm and the AEC to select mAs, they found that inclusion of the grid improved SDNR for all thicknesses tested. However, the improvement came with a substantial increase in radiation dose, ranging from a factor of 2 for 10 cm thickness to over a factor of 6 for 25 cm thickness. The corresponding SDNR increased by factors of 1.5 and 4.0, respectively.

The impact on radiation dose of an antiscatter grid used in conjunction with AEC depends critically on the programming of the particular AEC system. In the study reported here, manual techniques were used and the exposure parameters were kept constant for a given phantom thickness in order to isolate the impact on image quality of the grid with radiation dose held fixed. The result was that addition of the grid

FIG. 10. (a) is a plot of the difference between grid-in and grid-out (grid-in-grid-out) values of (a) SD, (b) SDNR, and (c) contrast versus % glandularity for all signal block compositions in each phantom. Error bars on the plots are the standard deviation of five trials of the grid-in acquisitions.

improved SD and contrast (6.5 and 8.5 cm thicknesses) or left it essentially unchanged (4.5 cm thickness) for glandularity $\geq 30\%$ [Figs. 10(a) and 10(c)]. Figure 10(b) shows that inclusion of the grid raises SDNR for all glandularities and thicknesses tested, increasing SDNR for the 100% glandularity block by factors of 2.4 and 4.2 at 6.5 and 8.5 cm phantom thicknesses, respectively. As might be expected, for lower scatter (4.5 cm thickness) conditions, the magnitude of SDNR increased only slightly in the presence of the grid. Somewhat surprisingly, the amount of SDNR improvement decreased with increasing glandularity, becoming almost negligible for the 100/0 block [Figs. 9(c) and 10(b)].

Since the radiation dose was held fixed for all scans of a given phantom thickness, and a large portion of the scattered radiation, as well as some of the primary, was removed by the grid, the total number of detected photons in the grid-

in projection images is lower than for the grid-out images. This reduction in photon count makes grid-in images noisier (Fig. 11). With increasing phantom thickness, the variation in voxel value in the posterior-to-anterior dimension (going from the top of the image toward the bottom in Figs. 5–7) increases.

Cupping artifacts in which higher μ values are observed at the periphery compared to central regions are characteristic effects of scatter in CT observed in transaxial slices. In the coronal DBT slices of this study, rather than an increase in voxel value around the skin line, there is a drop seen in estimated attenuation around the periphery of the breast phantom of the DMT grid-out reconstructions. Lower voxel values around the periphery of the breast were also observed in DBT scans of a similar (Model 012A) CIRS phantom by Liu and Li, which they explain are due to an increase in scatter intensity in that region.²³ There is also a

FIG. 11. Plot of the standard deviation σ_B of the voxels within the background ROIs above the 50/50 block for each tested phantom thickness. Grid-out noise is represented by triangles connected by a dashed line, and grid-in noise is plotted using circles connected by a solid line. Error bars from the standard deviation of five trials of the grid-in acquisitions are included but are smaller than the plotted circular symbols.

fat-simulating (0/100 glandular/adipose) layer that surrounds the 011A phantom resulting in a reduction in attenuation at the outer-most edges relative to that in the background (50/50) region of the phantom. This composition change results in a true reduction in voxel values at the extreme periphery. Unlike CTBI, where the breast is more cylindrically shaped, the breast is compressed in DMT scans by a compression paddle and takes on a flattened shape, like the 011A phantom. The compressed shape gives rise to larger SPR values near the anterior periphery of the breast, as reported by Sechopoulos *et al.*²⁴ They found that scatter arising from the regions of the breast support and detector cover outside the breast region can fall beneath the peripheral portion of the breast not in contact with the breast support and raise the SPR in that region relative to more central regions.²⁴ Thus, while the presence of scatter always results in artifactual reduction in the estimated μ values, the spatial dependence of the voxel value nonuniformity is partially a function of the spatial distribution

TABLE III. Linear fit parameters for calculating image-based attenuation coefficients.

Tube voltage (kV)	Grid-in		Grid-out	
	m (voxel value/cm ⁻¹)	k (voxel value)	m (voxel value/cm ⁻¹)	k (voxel value)
28	6688	613	6368	881
29	6720	617	6416	856
31	6788	630	6483	828

of the scatter radiation and thus on the particular breast (or phantom) shape.²⁵

A possible reason for the larger improvement of SDNR and relative accuracy for the more glandular blocks compared to the blocks of lower glandularity [Fig. 10(b)] is a difference in the SPR among the blocks. Monte Carlo simulations performed by Boone *et al.* demonstrated that the SPR changed somewhat with glandularity, but not in a noticeable trend.²⁶ In contrast, Kwan *et al.* examined the dependence of SPR on glandularity in CTBI and found that SPR increased with increasing glandularity at the center of a 14 cm diameter phantom.²⁷ Shen *et al.* found the same trend with a slot-scanning digital mammography system where the SPR for the 8 cm phantom simulating a breast composition of 100% adipose tissue was 0.16 as opposed to a SPR of 0.19 for the 100% glandular phantom of the same thickness.²⁸ Similarly, Fig. 4(a) shows an increase in grid-out SPR with increasing glandularity, while grid-in SPRs for different compositions are nearly identical. Thus, scatter removal by a grid reduces the SPR by a larger fraction for higher glandularity blocks than for the lower glandularity blocks.

Difference in SPR may also account for why the correction by the grid of the estimated μ values for the 8.5 cm phantom is slightly lower than for the 6.5 cm phantom. In Fig. 4(b), the SPR of the 8 cm PMMA phantom in the presence of the W-poly grid is slightly higher than that of the 6 cm phantom. Even though grid-out SPR for 8 cm is higher at 1.36 when compared to 0.87 for the 6 cm phantom, the resulting scatter

FIG. 12. Plots of mean voxel value versus theoretical attenuation coefficients of material with 0/100, 30/70, and 50/50, 70/30, and 100/0 compositions acquired at tube voltages of (a) 28 kV, (b) 29 kV, and (c) 31 kV.

FIG. 13. Linear attenuation coefficients of embedded 1 cm blocks derived from image voxel values and the linear fit parameters of Table III. Plots show the calculated values of μ_S versus glandularity for the grid-in images (circles with solid line), the grid-out images (dashed line with triangles), and the theoretical values μ_T (cross symbols connected with solid lines) for the (a) 4.5 cm, (b) 6.5 cm, and (c) 8.5 cm phantoms.

rejection by the grid cuts down the SPR by a factor of 5.8 for the 8 cm phantom and by a factor of 6 for the 6 cm phantom. Since a larger fraction of scatter was removed in the 6.5 cm case, the correction in μ by the grid would be slightly better than for the 8.5 cm phantom.

The observed percent errors in image-based μ of 5% or more compared to theoretical values can be attributed partially to the polyenergetic beam and partially to DBT artifacts resulting from its incomplete angular sampling. The incomplete sampling results in the smearing of attenuation in the direction perpendicular to the planes of the reconstructed slices. Thus, unlike CT, in which voxel values are representative of the attenuation in a small volume centered on each voxel, in DBT, they are impacted by the attenuation in slices above or below the voxel in question. Figure 13 shows that, in the presence of scatter, the μ values of blocks with glandularities lower than those of the surrounding 50/50 background are artificially raised, while those with higher glandularity are lowered. This finding is consistent with the fact that slices above and below the block slices contain 50/50 material. Additionally, depending upon the choice of

reconstruction algorithm, voxel values can also be affected by neighboring structures within the same slice. Zhang *et al.* calculated a line object spread function (LOSF) from DBT images of a wire phantom that were reconstructed using three different reconstruction algorithms: backprojection (BP) method, the simultaneous algebraic reconstruction technique (SART), and the maximum likelihood method with the convex algorithm (ML-convex).³ They found that edge enhancement effects, similar to the results of unsharp mask filtering, were observed when using the iterative SART and ML-convex algorithms, which they explain are due to the estimates of the average differences in μ in the backprojection step of the reconstruction.³ As a result, any edge enhancement could also impact voxel values of adjacent structures.

The variations in SD, SDNR, and contrast measured over five trials are small for both the 70/30 and 100/0 blocks, but slightly larger for the 70/30 blocks. The fact that the intervals within the error bars in Fig. 10(b) all lie well above zero for both mass-simulating signal blocks for the 6.5 and 8.5 cm thick phantoms provides confidence that the grid improves SD, SDNR, and contrast for those thicknesses.

FIG. 14. Percent error in image-based estimates of attenuation coefficients as compared to theoretical values μ_T plotted versus glandularity for the (a) 4.5 cm, (b) 6.5 cm, and (c) 8.5 cm phantoms. Grid-in results (solid lines and circles) and grid-out results (dashed lines and triangles) are shown.

5. CONCLUSIONS

Suppression of voxel values, and thus in estimates of linear attenuation coefficients caused by the presence of scatter, not only decreases lesion contrast but also can even potentially cause subtle masses to blend in with healthy background tissue as observed by the near zero or even negative SD values of the 70/30 block [Fig. 9(b)]. Unlike FFDM, DBT provides some estimate of local μ values. However, this study showed that despite effective scatter rejection, there were still residual errors in image-based estimates of μ due to DBT's limited angular sampling range and the associated incompleteness of the projection data. The fact that even after scatter rejection estimates of μ are too high for objects with attenuation lower than the surrounding tissue and too low for objects whose attenuation exceeds that of the background (Figs. 13 and 14) suggests that voxel values in a given slice are changed according to the relative attenuations of objects in neighboring slices. Although it was beyond the scope of this study, it is likely that larger acquisition angular range could further reduce the residual errors.

Our initial experience with the prototype 2D focused grid used with the investigational DMT scanner shows that DBT scans with isocentric systems can benefit from the addition of an antiscatter grid. For the range of phantom thicknesses tested here, the SD, SDNR, and lesion contrast all either improved or remained approximately unchanged with the addition of the grid under conditions of fixed radiation dose. Therefore, a compensating increase in radiation dose might not be necessary if such a grid were used for DBT scans of patients whose compressed breast thickness falls within the range tested here (4.5–8.5 cm).

Quantitative measurement of SD, contrast, and SDNR at low spatial frequency, along with subjective determination of visibility of graded-size, high- and low-contrast objects such as those in the ACR accreditation and the CIRS Model 011A phantom are useful methods for initial assessment of the impact of the grid on image quality. However, future studies incorporating phantoms designed for 3D imaging, and less subjective spatial frequency dependent metrics, such as the task-specific detectability index, would provide a more complete characterization. Ultimately, observer studies incorporating human or mathematical observers and physical or mathematical phantoms with nonuniform backgrounds are needed for a realistic assessment of other potentially clinically important effects. These include the impact on lesion detectability after removal by the grid of low-frequency, scatter-induced, location-dependent artifacts such as those indicated in the profiles of Figs. 7 and 8, as well as the impact of any residual grid-shadowing artifacts.

In this study, SDNR for the 4.5 cm breast phantom was unchanged with inclusion of the grid. The decreasing SDNR improvement with decreasing breast thickness suggests that for thin breasts the improvement in SD afforded by removal of scatter radiation by the grid could be outweighed by the removal of primary radiation. On the other hand, in DBT scans of larger, mostly fatty breasts where dose is typically raised to overcome the effects of scatter,²⁹ the required dose

may be lower for tomosynthesis acquisitions if scatter removal techniques were available.

Tomosynthesis is now becoming a new clinical standard, but there are currently no methods of alleviating the negative effects of scattered radiation on DBT image quality for a vast majority of clinical DBT systems. A better quality DBT scan could be possible through the use of grids such as the one tested here.

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^{a)}Author to whom correspondence should be addressed. Electronic mail: tp3m@virginia.edu

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