Comparison of breast specific gamma imaging and molecular breast tomosynthesis in breast cancer detection: Evaluation in phantoms

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Purpose: Breast specific gamma imaging or molecular breast imaging (BSGI) obtains 2D images of ^{99m}Tc sestamibi distribution in the breast. Molecular breast tomosynthesis (MBT) maps the tracer distribution in 3D by acquiring multiple projections over a limited angular range. Here, the authors compare the performance of the two technologies in terms of spatial resolution, lesion contrast, and contrast-to-noise ratio (CNR) in phantom studies under conditions of clinically relevant sestamibi dose and imaging time.

Methods: The systems tested were a Dilon 6800 and a MBT prototype developed at the University of Virginia. Both systems comprise a pixelated sodium iodide scintillator, an array of position sensitive photomultipliers, and a parallel hole collimator. The active areas and energy resolution of the systems are similar. System sensitivity, spatial resolution, lesion contrast, and CNR were measured using a Petri dish, a point source phantom, and a breast phantom containing simulated lesions at two depths, respectively. A single BSGI projection was acquired. Five MBT projections were acquired over $\pm 20^\circ$. For both modalities, the total scan count density was comparable to that observed for each in typical 10 min human scans following injection of 22 mCi (814 MBq) of ^{99m}Tc -sestamibi. To assess the impact of reducing the tracer dose, the pixel counts of projection images were later binomially subsampled by a factor of 2 to give images corresponding to an injected activity of approximately 11 mCi (407 MBq). Both unprocessed (pixelated) BSGI projections and interpolated (smoothed) BSGI images displayed by default on the Dilon 6800 workstation were analyzed. Volumetric images were reconstructed from the MBT projections using a maximum likelihood expectation maximization algorithm and extracted slices were analyzed.

Results: Over a depth range of 1.5–7.5 cm, BSGI spatial resolution was 5.6–11.5 mm in unprocessed projections and 5.7–12.0 mm in interpolated images. Over the same range, the in-slice MBT spatial resolution was 6.7–9.4 mm. Lesion contrast was significantly improved with MBT relative to BSGI for five out of eight lesions imaged at either the 22 mCi or the 11 mCi dose level ($p < 0.05$). At both dose levels, significant improvements in CNR with MBT were also found for five out of eight lesions (9.8, 7.8, 6.2 mm lesions at water depth of 1.7 cm and 9.8, 7.8 mm lesions at water depth of 4.5 cm, $p < 0.05$). The 6.2 and 4.9 mm lesions located at 4.5 cm below the water surface were not visible in either modality at either activity level.

Conclusions: Under conditions of equal dose, imaging time and similar detectors, compared to BSGI, MBT provided higher lesion contrast, higher CNR, and spatial resolution that was less depth dependent. © 2015 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4922398>]

Key words: breast specific gamma imaging, molecular imaging, nuclear medicine, breast, tomosynthesis

1. INTRODUCTION

Mammography serves as the current clinical standard for early breast cancer detection. Although its sensitivity ranges from 71% to 96% when averaged over all breast types, the sensitivity decreases to 68% in women with radiographically dense breasts and 48% in women with extremely dense breasts.^{1,2} Unfortunately, not only does high breast radiodensity create a challenge in imaging but also is an independent risk factor for developing cancer. Women with radiodense breast tissue

have up to six times greater lifetime risk of being diagnosed with breast cancer than do those with nondense breast (fatty or scattered fibroglandular).^{3–6} In the US, over half of women under age 50 and one third of women over 50 have either heterogeneously dense or extremely dense breasts, amounting to ~40%–50% of the overall screening population.^{7,8}

Unlike mammography, where anatomical information is obtained through differences in the x-ray attenuation coefficient of tissues, breast specific gamma imaging (BSGI), otherwise known as molecular breast imaging (MBI), provides

physiological information by imaging a gamma-ray emitting tracer, usually ^{99m}Tc sestamibi, that preferentially accumulates in malignant cells. A recent retrospective study of 341 women with biopsy-proven breast cancer diagnosed from 2004 to 2009 suggests that the sensitivity of BSGI is similar in women with dense (94.7%) and nondense breasts (96.5%).⁹ The lack of correlation between BSGI performance and radiographic breast density can be attributed to the fact that unlike the lower photon energies used in mammography, transmission of the 140 keV gamma radiation is largely unaffected by differences in the attenuation coefficients of adipose and fibroglandular tissue at this energy.^{5,10}

Like breast MRI, BSGI is being considered as a screening modality for women at high risk for breast cancer.¹¹ In light of this, studies have been made comparing the performance of BSGI to that of contrast-enhanced MRI (CE-MRI) as adjuncts to x-ray mammography. In a study of 49 patients comparing contrast MRI with conventional, large-camera scintimammography, Imbriaco *et al.* found no statistically significant difference in either sensitivity or specificity.¹² However, using a dedicated camera to compare CE-MRI with BSGI among 33 women with indeterminate mammographic findings, Brem *et al.* found comparable sensitivity, but significantly greater specificity (71%) for BSGI relative to MRI.¹³ In a multicenter study of women with biopsy-proven invasive lobular carcinoma, Brem *et al.* compared the sensitivities of mammography, US, MRI, and BSGI among 26 women with a total of 28 biopsy-proven invasive lobular carcinomas.¹⁴ Using final surgical pathology results as the reference standard, the sensitivity of BSGI was 93%, whereas mammography, US, and MRI had sensitivities of 79%, 68%, and 83%, respectively. Although this study was too small to obtain statistically significant conclusions, these trends indicate a potentially valuable role for BSGI in detection of lobular carcinoma.

While BSGI obtains a single 2D projection image of the sestamibi distribution in the breast, molecular breast tomosynthesis (MBT) maps the tracer distribution in 3D.^{15–17} In BSGI, the breast is immobilized by a high attenuation plate and the gamma camera itself so that the breast to collimator distance is minimized for spatial resolution maximization. In MBT, the detector acquires multiple projection views over a limited angular range, with the breast under mild compression, similarly to BSGI. MBT, when added to digital breast tomosynthesis (DBT), has demonstrated improvements in specificity and positive predictive value compared to DBT alone in a pilot study.¹⁷ The objective of this study is to compare the performances of BSGI and MBT in phantom studies under conditions of clinically relevant dose and imaging time. Spatial resolution, lesion contrast, and contrast-to-noise ratio (CNR) were evaluated.

2. MATERIALS AND METHODS

2.A. Molecular breast imaging systems

The systems compared were a Dilon 6800 (BSGI) (Dilon diagnostics, Newport News, VA) and a MBT prototype used as the functional imaging part of the dual modality tomosynthesis (DMT) scanner built at UVA^{15–17} (Fig. 1). Both systems comprise a pixelated NaI(Tl) crystal coupled to position sensitive photomultiplier tubes (PSPMTs), and both use parallel hole collimation with a low-energy-high-sensitivity (LEHS) lead hexagonal hole collimator. The BSGI camera is mounted on an adjustable arm. The MBT camera is mounted on a gantry arm permitting both rotational and radial translational motions to vary both viewing angle and the axis of rotation (AOR)-to-detector distance. The specifications of the two systems are summarized in Table I.

FIG. 1. Dilon 6800 BSGI system (left) and MBT scanner (right). MBT is a subsystem of the DMT scanner. During a DMT scan, the patient is seated. X-ray DBT is performed with the gamma camera positioned anteriorly, out of the field of view. Following DBT, the camera is positioned at the chest wall and the MBT scan is performed. For clarity, the photo shows the gamma camera at the chest wall and positioned far above the paddle; however during MBT, the camera is lowered to just above the paddle as shown in Fig. 2.

TABLE I. Specifications of BSGI and MBT systems.

	Field of view (cm × cm)	Collimator hole length (mm)	Collimator hole diameter (mm)	Septal thickness (mm)	Collimator efficiency [(counts/min)/μCi]	Pixel size (mm)	Intrinsic resolution (mm)	Energy resolution (%)	Energy window (keV)	Overall camera sensitivity [(counts/min)/μCi]
BSGI	20 × 15	21.0	1.85	0.3	941 ^a	3.2	3.3	13.5	126–154	458
MBT	20 × 15	21.5	1.90	0.3	952 ^a	2.2	2.4	13.5	121–163	617

^aCollimator efficiency was calculated using Eq. (16-7) from Ref. 27: $g = K^2(d/l_e)^2[d^2/(d+r)^2]$ where $K = 0.26$ for hexagonal hole.

The major differences between the two systems are pixel size (equal to the crystal pitch) and energy window.^{15,18} In BSGI, the window is set at +10%/–10% of the 140 keV photopeak energy; whereas in MBT, it was set slightly wider to +16%/–14% to ensure that the maximum number of primary (nonscattered) photons was included in the images. The overall sensitivity of each camera was measured using the protocol specified by the National Electrical Manufacturers Association (NEMA).¹⁹ A thin layer of water with a known amount of ^{99m}Tc sestamibi was placed in a Petri dish and imaged for 2 min. The total counts received in a large region of interest (ROI) surrounding the dish image was measured and corrected for radioactive decay and background counts.

2.B. Image acquisition, processing, and reconstruction

Figure 2 compares the acquisition schemes of BSGI and MBT. During BSGI scans, the phantom was positioned directly against the collimator input surface and a single view was acquired. During MBT scans, the phantom was positioned between a compression paddle and a breast support, with the support about 5 cm below the camera's AOR. In BSGI, compressed thickness, T , was defined as the separation between the collimator input surface and the high attenuation plate used to immobilize the breast; lesion (or source) depth

was defined from the collimator input surface. In MBT, compressed thickness, T , was defined by the separation between the compression paddle and the support; lesion (or source) depth was defined from the lower surface of the paddle. Five MBT projections were acquired over $\pm 20^\circ$ with a fixed 10° step. For each view, the camera was positioned radially so as to minimize the collimator-to-compression paddle distance. For a breast with compressed thickness of 7.3 cm, the AOR-to-collimator distance ranged from 4.2 cm for the zero-degree view to 8.3 cm at the extreme ($\pm 20^\circ$) views.

In clinical BSGI, interpolated (smoothed) images with larger matrix size are presented by default to radiologist who can also opt to read unprocessed (one pixel per detector element) projections if desired. In this study, both unprocessed projections and the standard interpolated images from the 6800 workstation were analyzed. For MBT, the five projection views used to reconstruct a volumetric image (containing slices as illustrated by the dashed lines in Fig. 2) using a maximum likelihood expectation maximization (ML-EM) algorithm.²⁰ In this study, no attenuation or scatter corrections were applied. An empirically determined Gaussian blurring kernel with a full width at half maximum (FWHM) corresponding to a 7 cm source to collimator distance (approximately equal to the average AOR to collimator distance in a MBT scan of a 7.3 cm thick breast) was used to perform resolution recovery. The reconstructions were performed on a Geforce GTX 680 graphic

FIG. 2. Schematics of BSGI and MBT acquisition. In BSGI, the breast is compressed between the camera and a high attenuation plate. In MBT, the breast is compressed between a breast support and a low-attenuation compression paddle. In this study, lesion (or source) depth was defined relative to the collimator input surface in BSGI and relative to the lower surface of the compression paddle in MBT.

FIG. 3. Schematic of point source phantom (a). Schematic of lesion size and location within the box phantom (b). The water surface in the box phantom was 5 mm below either the collimator in BSGI or the lower surface of the compression paddle in MBT. The water height was 7.2 cm. The dashed rectangular region represents background ROI drawn on either BSGI projections or MBT slices.

processing unit with seven iterations, requiring a total of 4 s. Aside from correction of the projection images for detector sensitivity and nonuniformity, no prior- or post-processing was performed on the projection images or the reconstructed slices.

2.C. Spatial resolution characterization

To characterize the spatial resolution of the MBT camera, a capillary source (<1 mm inner diameter) was placed at distances 1–8 cm from the collimator surface with a 1 cm increment. Multiple one-dimensional (1D) profiles across the line source in the projection image were averaged and fitted with Gaussian profiles. The FWHM of the fitted profiles was reported at each source-to-collimator distance.

A phantom was imaged using a BSGI scan and then a MBT scan. It was comprised of seven equally spaced acrylic posts whose heights varied linearly with post position [Fig. 3(a)]. The tops of the shortest and longest posts were 0.5 and 6.5 cm above the 0.5 cm thick baseplate, respectively. Tiny hemispherical wells (<1 mm diameter) were drilled on the post tops and filled with $^{99\text{m}}\text{Tc}$ solution to create point-like sources. With the phantom placed on the breast support or the high attenuation plate and T adjusted to a value of 8.5 cm, the point sources were located at depths ranging from 1.5 to 7.5 cm.

Using ImageJ (NIH, Bethesda, MD), 1D profiles centered on the point sources in the BSGI projection were extracted and fitted with Gaussians. For MBT, the 1D profiles were extracted from slices parallel to the breast support and centered on each source.

The FWHM of each Gaussian profile was calculated using the following equation:

$$\text{FWHM} = 2\sqrt{2 \ln(2)} \cdot \sigma, \quad (1)$$

where sigma is the standard deviation of the fitted Gaussian profile. For both BSGI and MBT, FWHM measurements in the x and y directions were averaged. In MBT, the x and y dimensions (in-slice directions) are defined within the extracted slices, and the z dimension is the depth direction. For both modalities, the averaged FWHM values were plotted versus source depth and a linear least-squares fit was applied.

2.D. Lesion contrast and CNR

A $16.5 \times 12.5 \times 7.7$ cm (inner dimensions) acrylic box phantom containing $^{99\text{m}}\text{Tc}$ water solution and eight hollow thin-walled acrylic spherical lesions was used [Fig. 3(b)]. The two groups of four lesions were mounted at different heights, with the centers of one group located 1.7 cm and the other 4.5 cm below the water surface when the phantom was filled with $^{99\text{m}}\text{Tc}$ solution to a depth of 7.2 cm. The lesion depths as defined in Sec. 2.B were thus 2.2 and 5.0 cm. The internal diameters of the four lesions in each group were 9.8 mm (0.5 ml volume), 7.8 mm (250 μl), 6.2 mm (125 μl), and 4.9 mm (63 μl). Each lesion group was supported by an acrylic post attached to the bottom of the phantom. The lesions were attached to the ends of two crossed, low attenuation rods that were mounted onto the top of each post. This phantom has been described previously.¹⁵ The thickness of the phantom was chosen to be equal to the average compressed breast thickness observed in a previous DMT human study¹⁵ (7.7 cm). In that study, the compression force was based on the technologist's assessment of the minimum force necessary to maintain

TABLE II. ROI sizes in BSGI projections and MBT slices.

Image type	Pixel size (mm)	Diameter of lesion ROI (mm)				Background ROI on the original output (pixel size: number of pixels)
		For 9.8 mm lesion	For 7.8 mm lesion	For 6.2 mm lesion	For 4.9 mm lesion	
BSGI: unprocessed	0.2 ^a	9.8	7.8	6.2	5.0	3.2 mm: 578 pixels
BSGI: interpolated	0.67	10.0	8.0	6.0	4.8	0.67 mm: 13 770 pixels
MBT slices	0.2 ^a	9.8	7.8	6.2	5.0	2.2 mm: 1200 pixels

^aImage resizing was performed.

FIG. 4. BSGI images and MBT slices of the point source phantom. The zero-degree projection view of the MBT scan is also presented.

proper positioning and immobility along with interaction with the subject regarding whether they would be comfortable for a 10 min scan. A retrospective analysis showed that on average, the compressed thickness was $(17 \pm 11)\%$ greater during DMT than for the same breast during the preceding clinical mammogram. A 20:1 lesion-to-background activity concentration ra-

tio was used based on the estimates from planar MBI studies performed at the Mayo Clinic.^{21,22} To assess statistical variation, eight identical trials of BSGI and MBT phantom scans were performed, with correction for isotope decay between repeated acquisitions. Lesion contrast and CNR were used to quantify lesion visibility and are defined as follows:

$$\text{Contrast} = \frac{\text{average lesion pixel counts} - \text{average background pixel counts}}{\text{average background pixel counts}}, \quad (2)$$

$$\text{Noise} = \frac{\text{standard deviation of background pixel counts}}{\text{average background pixel counts}}, \quad (3)$$

$$\text{CNR} = \frac{\text{contrast}}{\text{noise}}. \quad (4)$$

Lesion counts were obtained from circular ROIs with diameters matching the known internal diameters of each sphere and centered on each lesion. For MBT, ROIs were drawn on the reconstructed slices centered on each lesion. To maintain roughly circular ROI shape and thereby minimize variability in the measurement of the average lesion pixel values, the unprocessed BSGI projections were resized from 80×80 (pixel size = 3.2 mm) to 1280×1280 (pixel size = 0.2 mm) matrix size using pixel replication in ImageJ. The MBT slices were resized from 90×66 (pixel size = 2.2 mm) to 990×726 (pixel size = 0.2 mm) using the same method. Since background ROIs occupied a much larger region as illustrated in Fig. 3(b), the original unprocessed projections or reconstructed slices were used for determination of background average pixel value and standard deviation. The interpolated BSGI images from the Dilon 6800 workstation had a matrix size of 384×384 (pixel size = 0.67 mm), so no resizing was

performed on them prior to lesion and background measurements. Table II summarizes the sizes of lesion and background ROIs.

2.E. Count density calibration

The injected ^{99m}Tc -sestamibi (marketed as MiralumaTM) activity currently recommended for breast imaging by the Society of Nuclear Medicine and indicated on the MiralumaTM package insert is 20–25 mCi (740–925 MBq), although recent studies with two-camera systems have demonstrated that substantially lower activity is possible.^{22,24} To determine a clinically relevant target count density for the projection images of the box phantom, 25 projection images (10 LMLO views, 10 RMLO views, 3 LCC view, 2 RCC views) of 10 subjects in a recent clinical trial of MBT were analyzed. In that study, subjects received an average of 22 mCi (814 MBq) of

FIG. 5. Spatial resolution in BSGI images and MBT reconstructed slices. The spatial resolution of the MBT camera is plotted and labeled as “MBT camera.” For the MBT camera, the depths were measured from the collimator input surface.

^{99m}Tc -sestamibi and a total acquisition time of 10 min was used. To determine the average projection view count density, the mean number of counts per cm^2 was measured in a large ROI drawn within the breast (avoiding the breast periphery) on the zero-degree projection view in each case. A linear least-square fit was applied to the measured mean pixel values plotted versus the compressed breast thickness. The fit was used to select the target count density for a given phantom thickness. The acquisition time necessary to achieve the target count density for the zero-degree view was also used for the other MBT views. The total BSGI acquisition time was set equal to that of the MBT scan with decay correction between scans.

FIG. 6. Count density in human MBT projection images (zero-degree view) vs compressed breast thickness, under conditions of 22 mCi injected ^{99m}Tc -sestamibi and 2 min acquisition time per projection view.

To assess the impact of reducing the tracer dose by half (11 mCi), the pixel counts of projection images were later binomially subsampled by a factor of 2 to give an image with half the count density of the original image, but retaining Poisson statistics. The reconstructed MBT slices or interpolated BSGI images derived from those subsampled projections were analyzed in the same way as described in Sec. 2.D.

3. RESULTS

3.A. Spatial resolution characterization

Point source phantom BSGI images and MBT slices at three depths are shown in Fig. 4. The zero-degree projection

FIG. 7. BSGI projection images and MBT reconstructed slices of the box phantom. Lesion-to-background concentration ratio was $\sim 20:1$. MBT slices through the lesion centers at depths of 2.2 cm (1.7 cm below the water surface) and 5.0 cm (4.5 cm below the water surface) are shown in the third and fourth columns, respectively. Images in the first row were produced from the original projections in which the count density is about equal to that in a 22 mCi, 10 min scan of a 7.2 cm thick breast. Those in the bottom row were produced from subsampled projections in which the count density corresponds to an 11 mCi, 10 min scan.

view of the MBT scan is also presented. In the BSGI images, all sources are visible. MBT slices, although they are 2.2 mm thick, contain signal from 4 to 5 sources because of the poorer spatial resolution in the depth direction compared to the in-slice resolutions due to the limited angular range (40°) used in this study.

The measured FWHM spatial resolutions of BSGI images and MBT slices are plotted versus source depth in Fig. 5. The FWHM planar spatial resolution of the MBT camera is plotted and labeled as “MBT camera.” For this planar resolution measurement, source depths were relative to the collimator input surface, defined in the same way as in a BSGI scan. The difference in spatial resolution between the unprocessed and interpolated BSGI images was minimal. Compared to BSGI, MBT has a slightly better spatial resolution in terms of the planar resolution of the gamma camera. The resolution in the reconstructed MBT slices was less location dependent than that in either the MBT projection images or BSGI images. Over a depth range of 1.5–7.5 cm, MBT FWHM spatial resolution was between 5.1 and 10.4 mm in planar imaging and between 6.7 and 9.4 mm in slices (tomosynthesis imaging), compared to 5.6–11.3 mm FWHM spatial resolution for BSGI in unprocessed projections and 5.7–12.0 mm FWHM in interpolated images.

3.B. Count density calibration

The count density in the zero-degree projection view of MBT scans of human subjects obtained under conditions of 22 mCi ^{99m}Tc -sestamibi and a 10 min scan is plotted versus compressed breast thickness in Fig. 6. The linear fit to the data suggests an inverse relationship between the count density and compressed thickness, with a median count density of 550 counts/cm² over a thickness range of 6.9–10.6 cm.

Since the water height of the box phantom was 7.2 cm, the acquisition time per view in MBT was selected to reflect the human study-based count density for a 7.2 cm compressed breast thickness. At this thickness, the fitted count density indicates a count density of 719 counts/cm². The measured count density in the zero-degree projection view of the MBT scan of the box phantom was 720 counts/cm². The count density per minute in the MBT scans should be about 1.35 times that of the BSGI scans since the overall sensitivity of the MBT camera is 1.35 times that of the BSGI camera, as shown in Table II. The measured count density in the 10 min BSGI projection of the phantom was 2600 counts/cm², giving a count density per minute of 260 (counts/cm²)/min. The count density per minute in the zero-degree MBT view of the same phantom was (720 counts/cm²)/(2 min) = 360 (counts/cm²)/min. The

FIG. 8. Lesion contrast in BSGI images and MBT reconstructed slices of the box phantom with count densities corresponding to 22 and 11 mCi of injected ^{99m}Tc -sestamibi and 10 min total scan time. Lesions whose diameters are labeled with a single quotation mark were located at 2.2 cm depth (1.7 cm below the water surface) and those labeled with a double quotation mark were located at 5.0 cm depth (4.5 cm below the water surface). Error bars represent 95% confidence intervals of the measurements over eight nominally identical trials. Statistically significant ($p < 0.05$) improved contrast relative to that of BSGI is noted with an asterisk.

FIG. 9. Lesion CNR in BSGI images and MBT reconstructed slices of the box phantom with count densities corresponding to 22 and 11 mCi of injected ^{99m}Tc -sestamibi and 10 min total scan time. Lesions whose diameters are labeled with a single quotation mark were located at 2.2 cm depth (1.7 cm below the water surface) and those labeled with a double quotation mark were located at 5.0 cm depth (4.5 cm below the water surface). Error bars represent 95% confidence intervals of the measurements over eight nominally identical trials. Statistically significant ($p < 0.05$) improved CNR relative to that of BSGI is noted with an asterisk.

ratio of the count densities per minute of the two systems (MBT/BSGI) was thus 1.38, in line with the camera sensitivity ratio.

3.C. Lesion contrast and CNR

Figure 7 compares BSGI projections and reconstructed MBT slices of the box phantom under the two simulated dose levels. Unlike BSGI where all lesions were visible in a single view, in MBT, lesions were visible in different slices depending on their depths. Although eight lesions were imaged, neither of the smallest two lesions (4.9 and 6.2 mm) was visibly identifiable in any image under any dose condition at 5.0 cm depth (4.5 cm below the water surface).

Figures 8 and 9 contain bar plots of lesion contrast and CNR measured and averaged over the eight nominally identical trials. The error bars represent 95% confidence intervals of the measurements for each imaging and processing condition. In addition, values were compared between BSGI and MBT using an unpaired two-tailed t -test. An alpha level of 0.05 was used to determine if changes in contrast or CNR with MBT relative to BSGI were significant. Because BSGI images can be presented in either unprocessed or interpolated form, the

comparisons with MBT were done for each form, respectively. For each lesion size and depth, the higher p -value among the two comparisons with MBT (MBT in-slice vs BSGI unprocessed and MBT in-slice vs BSGI interpolated) was selected and compared against the 0.05 significance level. Because the null hypothesis is that there is no difference between MBT and BSGI in contrast and CNR, a higher p -value makes the null hypothesis harder to reject and the alternative (MBT is different than BSGI) harder to accept.

As expected, lesions located deeper or smaller in size have less contrast and CNR for both systems. Also as expected, lesion contrast remained at about the same level when the activity was reduced by half. Out of eight lesions imaged, significantly improved contrasts compared to BSGI were observed with MBT for five lesions at either the 22 mCi or the 11 mCi level. Significant improvements in CNR with MBT were also found for five out of eight lesions at both dose levels. The difference between MBT and BSGI in terms of CNR for the 4.9 cm lesion at 2.2 cm depth (1.7 cm below the water surface) was not significant. The 4.9 mm lesion and 6.2 cm lesion at 5.0 cm depth (4.5 cm below the water surface) were probably at the detection limits of both systems so the differences were also not significant.

4. DISCUSSION AND CONCLUSION

This study compared single-camera tomosynthesis in molecular breast imaging with the standard single-camera planar technique. The focus was on the relative performances of the two acquisition strategies using similar cameras, similar flat compression for breast immobilization, and under the same activity and acquisition time conditions. In the BSGI scans, counts were accumulated in a single view in the direction of breast compression where spatial resolution was maximized with the breast compressed directly against the camera. In MBT scans, counts were split over five views with the breast immobilized by a separate paddle-support structure.

The reconstructed in-slice MBT spatial resolution showed less depth dependence than in the projection images (either BSGI or MBT) due to the inclusion of resolution recovery during reconstruction. As seen in Figs. 4 and 7, with 40° angular range, MBT was able to resolve activity distribution in the depth dimension. Though the spatial resolution in this dimension was not comparable to that within a slice, it still permitted some degree of 3D lesion localization. In recent human studies that include both BSGI and MBT scanning of the same subject, we have found that even limited depth resolution can successfully reduce the amount of background tracer appearing in individual MBT slices compared to a full projection image, lowering background tracer superposition in a manner analogous to the reduction of normal tissue superposition in x-ray tomosynthesis compared to mammography.

The decreasing trend in count density with increasing breast thickness observed in prior human studies (shown in Fig. 6) could be attributable to several factors such as gamma ray attenuation within the breast, greater adipose fraction with increasing thickness, or simply lower average breast tracer concentration associated with the distribution of a fixed amount of tracer over a greater overall body mass. The linear fit used here may not be the best way to summarize the data but was employed in order to provide a rough empirical guideline for count density selection in the phantom studies.

The use of a slightly wider (+16%/–14%) energy window in MBT compared to the standard (+10%/–10%) window was meant primarily to ensure that essentially all events in the upper portion of the photopeak were captured. However, the lower minimum threshold may have allowed additional scatter counts in the image. The phantoms used in this study did not allow for simulation of scattered photons arising from the adjacent torso including heart and liver, which may reduce lesion contrast and CNR near the chest wall. However, using Monte Carlo simulation, Hruska and O'Connor found that for 6 cm compressed thickness, 16% of events acquired in the standard 126–154 keV (+10%/–10%) window of a camera with 15% FWHM energy resolution have undergone scattering, and that 21% of those originated in the torso (approximately 3.3% of all events in the window).²³ Thus, it is possible that the addition of scattered events from the torso would not make a material difference for the comparison performed here.

Parallel hole collimators whose hole shape and pitch are designed to match those of the detector elements can theo-

retically improve sensitivity compared to conventional unmatched (nonmatching shape or pitch) collimators if they are precisely aligned so that the collimator septa overlies the inactive areas between detector elements.^{25,26} The amount of sensitivity increase possible through the use of matched collimators depends on the open area fraction of the crystal array, the accuracy of collimator and crystal array fabrication, and precision of alignment. The amount of possible improvement is reduced as the divergence of the gamma rays exiting the collimator increases (i.e., with decreasing hole length-to-diameter ratio) and with increasing gap size between the collimator exit surface and detector entrance surface. In this study, unmatched collimators were used in both systems; however matched collimators could be incorporated into either.

Commercially available double-sided planar MBI systems (GE Healthcare, Gamma Medica) use two opposing cadmium zinc telluride (CZT) detectors to obtain 2D projection images from opposite sides of the breast.^{22,24} With this geometry, an effective upper limit for the collimator-to-lesion separation is half the compressed breast thickness. Reduction in the effective maximum lesion depth for each camera permits collimators with reduced spatial resolution and increased sensitivity (i.e., with smaller hole length-to-diameter ratios) to be utilized. A direct comparison of single camera MBT with double-sided MBI systems is beyond the scope of this study. It should be pointed out that double-sided MBT is entirely possible given proper gantry and breast immobilization system designs. Studies comparing double-sided planar MBI with both single-camera and double-sided MBT are ongoing.

In either one or two camera format, MBT provides depth information that cannot be naturally obtained by any planar technique. That information is valuable in lesion localization and biopsy and is essential for quantitative assessment of radiotracer concentration. The format of MBT images is also directly analogous to that of x-ray digital breast tomosynthesis and thus provides a useful way to form combined hybrid structural–functional image sets as part of an overall breast care program.

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