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Automated Breast Ultrasound: Dual-Sided Compared with Single-Sided Imaging

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Abstract

We report the design and performance of a mammographically configured, dual-sided, automated breast ultrasound (ABUS) 3D imaging system. Dual-sided imaging (superior and inferior) is compared with single-sided imaging to aid decisions on clinical implementation of the more complex, but potentially higher quality dual-sided imaging. Marked improvement in the image quality and coverage of the breast is found in dual-sided ultrasound over single-sided ultrasound. Among hypoechoic masses imaged, there are increases in the mean contrast-to-noise ratio (CNR) of 57% and 79%, respectively, for spliced dual sided vs. superior or inferior single-sided imaging. The fractional breast volume coverage, defined as the percentage volume in the transducer field of view that is imaged with clinically-acceptable quality, is improved from 59% in both superior and inferior single-sided imaging to 89% in dual-sided imaging. Applying acoustic coupling to the breast requires more effort or sophisticated methods in dual-sided imaging than in single-sided.

Keywords

Automated Breast Ultrasound; Mammography; 3D Imaging; Compression; Medial Lateral Oblique; Cephalo-caudal

Introduction

3D breast imaging has become an increasingly valuable tool for cancer screening and diagnosis (Kelly and Richwald 2011). In the U.S. and much of Europe, breast ultrasound is an adjunct to mammography or its 3D improvement, digital breast tomosynthesis (DBT). DBT improves breast cancer detection over mammography alone, but still misses many cancers that ultrasound will detect due to the similar mammographic density of cancers and many dense normal structures (Mariscotti, et al. 2014). Adjunctive ultrasound (US) screening has been shown to increase sensitivity over mammography, particularly in dense

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breasts (Leconte, et al. 2003). Breast cancer risk for women with the most dense breast tissue is four to six times greater than for women with fatty or no dense breast tissue (Boyd, et al. 2011). Younger women, with their higher potential for quality life years, have a higher proportion of dense breasts (Shen, et al. 2015) and they generally have more aggressive tumors. In one study (Webb, et al. 2014), over half of 609 patients dying from breast cancer were under age 50 at the time of their death. Cancers in dense breasts can be found more easily in ultrasound because of the relatively high ultrasound image contrast between cancer and most normal surrounding tissue and other ultrasound features of breast cancer.

Often with ultrasound screening, an additional diagnostic handheld exam (recall) is required. The financial and psychological costs of recalls for patients can be fairly high (Alcusk, et al. 2014, Harvey, et al. 2014). Several recent studies have demonstrated that the diagnostic power of automated breast ultrasound (ABUS) is comparable to that of handheld imaging. One such study reported a diagnostic accuracy of 97.7% for ABUS and 96.5% for hand scanning Choi, et al. 2014). In another study, analysis of ABUS images and images from hand-scanning led a perfect match-up in the masses classified at BI-RADS categories 4 and 5. (Kim, et al. 2013). Two other recent studies have investigated the image quality of masses in ABUS images compared with those produced by hand scanning (An et al. 2015, Kuzmiak et al. 2015). ABUS is desirable because of its efficiency (Kaplan 2014) and orderly coverage for comparison with previous, similar exams. The results of other studies are not as optimistic. In one recent study, three readers with hand-held ultrasound experience between six and twelve years and ABUS experience between none and six months achieved cancer detection sensitivities of only 57.1% to 78.6% with an ABUS system (Chang, et al. 2011). In another study, a reader with four years experience in ultrasound achieved 100% sensitivity, but only 52.8% sensitivity, with an ABUS system (Wojcinski, et al. 2011). It is possible that extensive experience with ABUS is needed to increase the sensitivity and specificity, as indicated by another study reporting improvement in the second year of US screening (Weigert and Steenbergen 2015). In cases going to biopsy from hand-held scanning, most cancers missed by ABUS were evident retrospectively. If ABUS and mammography or DBT images are acquired in the same or similar compression, then the images can be viewed side-by-side. This could make the analysis easier, possibly increasing sensitivity and decreasing callback rates. Combined systems for this purpose have been studied (Besson and Nields, Dines, et al., Dines 1995, Kapur, et al. 2004, Padilla, et al. 2013, Richter 1997, Schmidt, et al. 2006, Schutze 1998, Shmulewitz, 1995, Shmulewitz, 1996, Smith 2014, Suri, et al. 2005) and improved to increase image quality, breast stability between the compression paddles and the effectiveness of acoustic gel coupling (Booi, et al. 2007, Kapur, Thomenius, LeCarpentier, et al. 2014, Li, et al. 2010, Sinha, et al. 2007).

Imaging at the high contrast and resolution of high-frequency ultrasound (8 to 12 MHz) has the drawback of faster beam attenuation, resulting in less penetration. For thick breasts, this can be especially problematic. The penetration problem can be circumvented by imaging from both sides of the breast, so that areas that could not be imaged well by one probe will be imaged well by the other. While transmission imaging in the mammographic compression has been demonstrated (Green and Taenzer), this is the first implementation of dual-sided pulse echo imaging in the mammographic geometry. We describe our dual-sided,

mammographically-configured automated breast ultrasound system and present preliminary results on its effectiveness in improving image quality and coverage of the breast.

Materials and Methods

A. Scanning System

i. Physical Setup—An Instrumentarium Performa mammography system (GE Healthcare, Milwaukee, WI, USA) is modified to hold two sets of Parker, 0.5 micron resolution Cartesian X-X' Y positioning systems (FPS-3XXX081309, Parker Hannifin Corp, Motion & Control Div., Troy, MI, USA) above and below compression paddles. Each paddle consists of a frame, over which a polyester mesh is stretched. Two probes (M12L, GE Healthcare, Milwaukee, WI, USA) are rigidly mounted to the positioning systems and coupled to the breast with a viscous, but spreadable, coupling gel, Lithoclear (Sonotech, Inc., Bellingham, WA, USA). The gel is administered with 50 mL syringes. This is often supplemented with a gel roll contained in a polyester mesh cylinder to minimize gel sag and dripping. The gel passes through the mesh paddles in small quantities to speed coupling to the transducer and breast and minimize bubble trapping at the paddle-breast boundary. More effective and speedier methods of distributing and containing US coupling materials have been and continue to be ongoing efforts with reasonable interim solutions (Li, et al. 2010, Shmulewitz, 1996). Logiq 9 scanners (GE Healthcare, Milwaukee, WI, USA) operate the probes and a motor controller (6K4, Parker Hannifin Corp, Motion & Control Div., Troy, MI, USA) and a windows computer coordinate and synchronize probe movement and image acquisition. The probes are operated in compounding mode, with five images taken at different beam-steering angles. The imaging is performed at a frequency setting of 10 MHz, which is a compromise between a lower 9 MHz setting that might be optimum for coverage in 3D imaging of the breast and the 12 MHz setting expected on the aging Logiq 9 by radiologists accustomed to imaging the breast with hand scanning. The image field of view was set to 51 mm depth. As this system includes photoacoustic tomography capabilities, we refer to it as the Breast Light and Ultrasound Combined Imaging (BLUCI) system (Carson, et al. 2011, Sinha, et al. 2010), Figure 1A.

For each patient, the breast is compressed between the two mesh paddles and then coupling gel is distributed to fill air gaps. At each slice position an image is acquired with the first probe, and then with the second, after a short interval allowing for decay of the field from the first probe. The positioner motors are turned off during this time to minimize electronic noise artifacts. The probes are then moved simultaneously to the next acquisition position. In this manner the probes are swept across the breast from left to right, imaging every 0.8 mm. These probes have a limited scan width of 38.4 mm, so for larger breasts more than one sweep is required to cover the whole area. The first sweep is taken closest to the chest wall, with the possible second sweep shifted outwards from the chest wall by 28.4 mm, and the possible third sweep shifted outwards an additional 18.4 mm. These sweeps are labeled posterior, middle, or anterior (POS, MID, ANT), respectively. The POS and MID sweeps shown in Figure 1B. The shifts are made less than the image width to provide a minimum of 10 mm overlap between the images to facilitate image splicing.

ii. Control User Interface—A graphical user interface (GUI) is written in MATLAB to allow for flexible definition of scan modes and areas by technologists. After Logiq 9 ultrasound scanner initialization and scan position calibration, the expected coverage area is displayed along with all logged information from that day. The operator can load preset probe positions for different scanning protocols and modify them to accommodate varying breast sizes and positions. Any abnormal operation during a scan can be aborted by a physical emergency stop button or by a GUI emergency stop button. After a completed scan, the operator can choose to transfer and de-identify the acquired DICOM files.

B. Image Splicing

Each patient imaging session produces several image stacks. Throughout the following, we will use the words “top” and “bottom”, or more succinctly, “TOP” and “BOT” to signify generally superior and inferior views in cephalo-caudal or in medial lateral oblique orientations. For every sweep, there is an image stack from both the top probe and from the bottom probe. First, all partial images from the top are spliced together horizontally to form one contiguous top image (Figure 2 A–C). The two images from the bottom are likewise spliced together to form a complete bottom image (Figure 2 D–F). These two complete TOP and BOT image stacks can be viewed separately. The two stacks are also spliced vertically to produce a final complete TOPBOT image stack (Figure 2 G) for expedient review and maximum understanding of the breast anatomy. We use a normalized arctangent function as a weighting factor to create spliced areas that transition smoothly from one image to another. The splicing algorithm is as follows:

$$slope=5; \text{ overlap=length(overlap_pixels);} \quad (1)$$

$$invtan1=-\operatorname{atan}\left(\frac{slope}{\text{overlap}} * \left(\operatorname{linspace}\left(-\frac{\text{overlap}}{2}, \frac{\text{overlap}}{2}, \text{overlap}\right)\right)\right); \quad (2)$$

$$invtan1=\frac{1}{2} * \left(\frac{invtan1}{\max(invatan1)}+1\right); \quad (3)$$

Shadowing is often helpful in mass detection because it provides information about the tissue directly in front of the shadowed region. Seeing this effect from both sides of a shadowing structure is expected to be even more revealing, particularly in the separate TOP and BOT images. The combined TOPBOT images show best the tumor shape and extent, including what are often hidden distal borders and tissues.

When there is no breast motion between sweeps, the overlap distance for horizontal splicing is known. The distance for the vertical splicing is less certain. The gap between the two compression paddles is not always an accurate measure of the breast thickness because, the compressed breast forces the mesh to bulge, making the actual breast thickness greater than

the paddle gap. As the ultrasound transducers are set on the mesh at a height that will not cause patient discomfort and their mounts are not totally rigid, their vertical separation varies from scan to scan. Thus, the vertical splicing distance is often determined by observation of homologous regions in TOP and BOT images. Automated image-based registration methods (Meyer, et al. 1999, Narayanasamy, et al. 2009, Neemuchwala, et al. 2006, Rohling, et al. 1998) are still too resource intensive and unpredictable for routine use in this task, given the artifacts and other causes of modest image quality often present at these ultrasound frequencies and depths.

C. Image Viewing System

The benefit of working in the mammographic geometry is that these US images can be viewed side-by-side with digital mammograms or digital breast tomosynthesis (DBT) images. Volunteers were recruited to be imaged on both the BLUCI system and with a combined DBT and single-sided US (DBT/US) system. For each patient we acquired three images: BLUCI dual-sided US, single-sided US and DBT. We created a MATLAB image viewing interface for side-by-side viewing of the images produced on the two systems (Figure 3). A large, high-resolution monitor with color (Coronis UNITI 12MP Fusion LED Color Display, Barco, Beaverton, OR, USA) was used to provide more options in image fusion including use of multiple US imaging modes.

In the combined DBT/US system, the breast does not leave compression between the DBT imaging and the US imaging, so the single-sided US (Figures 3C and D) and DBT images are usually well-aligned and additional registration provides only a small correction. For the dual-sided US, however, registration is very important for achieving alignment to the DBT/US, since the breast may be positioned, rotated, or compressed differently on the standalone dual-sided system than on the DBT/US system. Approximate registration of the dual-sided US to the DBT and of the US in the combined system to the DBT are achieved through translations that match homologous points. Unfortunately, US-to-US image-based registration (Narayanasamy, et al. 2009) between the two separate systems has proven to be quite challenging. It is difficult to position and compress the breast in exactly the same way, and the induced displacement fields are not easily corrected with automated registration methods due to US noise and artifacts and the sometimes steep displacement gradients.

D. Human Studies

This study involving 25 human imaging exams was IRB approved for asymptomatic volunteer women and patients with masses. Informed consent was obtained from each participant. Of the 25 cases, eight had one or more lesions. Two had biopsy proven cancers, and our system was able to image one of them. The other cancer was likely in the field of view that our system covered, but could not be found upon examination of the images.

i. Contrast Measure of Hypoechoic Masses—To quantify the improvement in visibility of large masses we use the following contrast-to-noise ratio (CNR) measure: $(\bar{S}_{out} - \bar{S}_{in})/\text{noise}$

$$CNR = \frac{\bar{S}_{out} - \bar{S}_{in}}{\sqrt{\sigma_{out}^2 + \sigma_{in}^2}} \quad (4)$$

Here $\bar{S}_{out}$ and σ_{out} are the mean and standard deviation of the decompressed pixel value in a region surrounding the mass, and $\bar{S}_{in}$ and σ_{in} are the mean and standard deviation of the decompressed pixel value within the mass. This CNR, sometimes referred to as contrast-to-speckle ratio (Cobbold 2007), is used less commonly than CNR with only σ_{out} in the denominator. The inclusion of both σ_{out} and σ_{in} in the denominator tends to account for the fact that masses with internal structure will tend to blend in with the tissue background more than do homogeneous masses.

First, all 25 datasets are examined and large masses from those datasets are selected for processing. The decision of whether a mass is large enough to be included in the analysis is made visually rather than by exact measurement. 13 masses are selected in total, and subsequent measurement of the diameters of these masses yields a mean diameter of 12.9 mm, so we cite this as a rough cutoff size for inclusion in the analysis. We measure the diameter as the average of the maximum length of secant lines through the cyst in the x, y, and z directions. Only larger masses are included in the analysis both to increase the sensitivity to low amplitude clutter that can make artifactual signals in cysts and other very hypoechoic masses, and also to ensure that the masses are large enough to provide good statistics. The one cancer we imaged was quite small, and therefore it is excluded from the analysis. The exclusion of the cancer is justified by the fact that cancers are often seen by shadowing rather than by the cancer itself, and the CNR measure we are using only measures the visibility of the mass itself relative to its local background.

Images containing selected masses are rescaled to 0.008 cubic mm per voxel to make processing faster, and then the selected masses are segmented in preparation for analysis. In order to perform the CNR measure on a mass, the mass must first be segmented. For the segmentation, the Robust Statistics Segmenter plugin (Gao, et al. 2013) of the 3D Slicer image processing package (Fedorov, et al. 2012) is used. This plugin takes manually selected seed points as inputs and then segments by a boundary evolution based on local statistics. After the 3D segmentation of a mass, the relevant regions for the contrast measure are extracted. An inner volume is used rather than the whole mass so that partial volume effects from boundaries are minimized. This inner mass volume is determined by taking each point of the surface and moving inwards towards the center of the mass by 2 mm if the distance from the center to the surface point is greater than or equal to 7 mm, moving inwards only 1 mm otherwise. The result is a volume that is roughly inscribed at least 1 mm inside the mass. A size-dependent inscribing distance is used to help ensure that the inscribed volume is large enough to produce good statistics for the CNR measure. The volume surrounding the mass is taken as the volume that is outside the mass but inside a box containing the mass. These regions for the CNR measure are shown in Figure 4. The compression curve employed for our imaging is that closest to purely logarithmic, approximated here as the pixel value Y in decibels from the transducer signal image X given by:

$$Y = [20 * \log_{10}(X+1)] * 255/DR \quad (5)$$

Where DR is the decibel range of the imaging system, which for our system was 72 dB. In order to evaluate the quality of the acquired image data, the images are decompressed from decibels back to the original intensity values. The resulting decompressed images, which are used for the CNR measure, are given by:

$$X = 10^{((1/20 * DR/255 * Y)) - 1} \quad (6)$$

ii. Coverage Comparison for Single-Sided and Dual-Sided Imaging—Another measure that is used to compare dual-sided and single-sided imaging is the fractional coverage, which is defined here as the ratio of the tissue volume coverage at clinically-acceptable image quality to the volume of tissue that would be imaged if the same path were swept out by probes that could image perfectly to an arbitrarily long range. It should be emphasized that the definition of ‘clinically-acceptable’ used is subjective, specified by the readers rather than by a quantifiable measure of image quality. However, given that analysis is ultimately performed visually by a breast imaging radiologist, this is deemed acceptable. Regions of low contrast and/or resolution that prevented tissue characterization are not counted towards the volume covered. An example segmentation is shown in Figure 5. Segmentation of clinically-acceptable tissue is performed manually on each of 25 image sets by two of the authors (hereafter referred to as ‘readers’). A further criterion used to aid in the segmentation is that for any region of tissue to be counted as reasonably imaged, the reader is instructed to estimate that 7 mm cysts in the region would have a high chance of being diagnosed correctly.

If a region is not easily identifiable, it is likely that all of the space directly beneath that region will also not be easily identifiable because US image quality generally decreases with increasing depth. Therefore, to segment the volume of high-quality imaging it is deemed sufficient to segment the boundary surface dividing the proximal region in which tissue was identifiable from the unidentifiable deeper region.

For each image, the skin line is segmented with a bilinear spline from spline points manually selected using a MATLAB interface. The two readers then use a second MATLAB interface to segment the boundary surface between acceptable and unacceptable image quality in the TOP and BOT images for all image sets. The total volume coverage is the volume between the segmented boundary surface and the pre-segmented skin line, as shown in Figure 6.

Results

A. Mass Contrast in Single- and Dual-Sided Imaging

Figure 7 shows the relationship between the CNR of a mass in single-sided imaging and the distance of that mass from the imaging transducer. The masses are split into two groups; one is simple cysts and the other is complex cysts and solid masses. Linear fits are shown for

both groups, and both fits have a negative CNR *vs. depth* slope, as expected. Information on the linear fits is given in Table 1. All CNR data, including the CNR of the masses in the TOPBOT images, is given in Table 2. For the CNR in the single-sided images, if the distance of the mass from the corresponding transducer is greater than 51.1 mm, then the mass is out of range of that transducer and would not be in the image even if the image quality were perfect at all depths, so the CNR is given as zero. The masses that were out of range were excluded from the plot in Figure 7, but they are included in Table 2. Many of the masses are only in the range of either the TOP or the BOT image, and not both. This demonstrates that dual-sided imaging is needed to capture all of the masses. One case where a mass imaged with the top probe is out of range from the bottom probe is shown in Figure 8A. However, even for the masses that are in range of both the top and the bottom probe, four out of five show a CNR that is significantly higher in the TOPBOT than in at least one of the single-sided images. One case where a mass is in range of both probes but only visible from the bottom probe is shown in Figure 8B. For most of the cases, the CNR of the mass in the TOPBOT image is very close to the larger of the CNR of the mass in the TOP or BOT images, implying that vertical splicing does not provide a significant improvement on the best single-sided image of the mass. Quantitatively, the percent increase in mean CNR from the TOP to the TOPBOT is 57%, and the percent increase in mean CNR from BOT to TOPBOT is 79%. In both cases the mean CNR is calculated including zero values corresponding to masses that are out of range of either transducer.

B. Fractional Coverage with Adequate Image Quality

In Table 3, means and standard deviations (σ) are given for the fractional coverage, inter-reader difference, and intra-reader difference. For the fractional coverage and the inter-reader difference, the statistics are done over 25 image sets. One reader repeats the measurements on four sets for inter-reader variation. On average, imaging from both the top and the bottom significantly increases the fractional coverage. The mean fractional coverage for imaging from the top only and from the bottom only were both 59%, while for imaging from both the top and the bottom it is 89% ($p < 10^{-12}$). A fairly large increase in fractional coverage is expected; at 10 MHz each probe can only image to down to around 4 cm at clinical image quality, and in this study the mean breast thickness in compression was 6.2 cm. The segmentation is shown to be reasonably objective by the relatively low inter- and intra-reader variability.

Discussion

One limitation of the study was the limited number of patients studied. A further limitation was the small number of cancers, which prompted a measure using larger cysts and solid masses instead. Also, acoustic coupling was not optimized to a rapid, bubble free application despite some efforts toward that end (Sinha, et al. 2007). Our dual-sided system requires coupling to both sides of the breast, which is more difficult than coupling to only one side would be. However, considerable progress has been made to make the coupling uniform and convenient. Since our results depend primarily on penetration, it should be emphasized that they only are valid for imaging at near 10 MHz. Imaging at a lower resolution would result in much better penetration, at the cost of a lower spatial resolution. The somewhat dated

Logiq 9 systems did not have an adjustment for mean speed of sound, nor effective transmit foci at all depths as is available now in some systems to improve resolution and signal level at the greatest imaging depths.

The expectation is that these dual-sided ultrasound systems will be used in tandem with standalone DBT and mammography systems, and also that dual-sided ultrasound eventually will be incorporated into DBT systems. Dual-sided ultrasound imaging might require nontrivial modification of some DBT systems to allow passage of a scanning ultrasound transducer, at least until the transducers can be made as a large, mammographically translucent plate. Whether the combined or separate systems approach becomes dominant, if either, will depend on the image quality of ultrasound achievable in the mammographic geometry and on the consistency with which breast positioning can be achieved in separate systems.

The CNR measure was targeted to large area contrast, minimizing effects of main lobe spatial resolution. One could add a third quantitative measure of such resolution using correlation lengths of speckle noise regions, but there were few areas of fully developed speckle in the breasts of these generally older patients. The subjective determination fractional volume coverage of the breast did include all obvious effects on image quality.

While not described here, our initial experience scanning the same patients on this dual-sided system as well as the combined single-sided ultrasound/DBT system suggest that it is difficult in some cases to obtain similar views of the breast in the separate systems. Improved methods of positioning the breast in the dual-sided ultrasound, while allowing for acoustic coupling, should improve consistency in positioning. Certainly, registration with DBT of the standalone dual-sided ultrasound in the mammographic geometry should be better than that achievable with automated scanning in the supine geometry. Speed of sound images would be especially helpful in DBT/US image based registration because the SOS contrast in common breast tissues is similar to that of x-ray contrast and is a measure of bulk properties of the tissues rather than of the tissue boundary properties. US-to-DBT, US-to-US and DBT-to-DBT image-based registration should improve tracking of low-risk abnormalities, cancer detection and use of nonionizing strategies to replace some x-ray imaging in younger women with dense breasts.

Automated elasticity imaging (Booi 2007, Booi, et al. 2007) and local compression by the scanning transducer (Carson, et al. 2010) also are possible with this type of system. Quasi-static elasticity studies were performed earlier (Booi 2007, Booi, et al. 2007) with solid, rather than mesh paddles. Compression of the entire breast by the paddles, to stabilize the breast position, provided preloading of all the tissues. This preloading resulted in an attendant increase in elastic modulus of the fat, reducing its contrast with breast cancer and most other solid masses. The elasticity imaging should show higher contrast when the minimum compression with mesh paddles is used and only local compression by the transducers or acoustic radiation force are employed to produce the required tissue strain for elasticity imaging. The mesh paddles require less compression of the breast to achieve stability because the frame compresses the retromammary tissues and the mesh bulges, locking the breast in place.

Conclusion

These studies suggest that a dual-sided ABUS system can increase both the contrast of masses (57% increase from top imaging only and 79% increase from bottom imaging only) and the fractional coverage (59% for imaging from the top or bottom only versus 89% for both), both of which are important for the detection of masses as well as for registration to other modalities. We expect that our ABUS system may have a role as a quick aid to diagnosis, however screening will benefit more strongly than diagnostic exams from whole breast US imaging and from registration with DBT, mammography, and other noninvasive modalities.

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Figure 1. Physical scan setup and process

(A) Physical scan system. (B) Diagram of scan setup showing breast under compression between the two mesh paddles. The blue arrows indicate the sweep direction of the probes,

which would be left to right from the point of view of the patient. The volumes corresponding to two different overlapping sweeps are shown in red and green. For the cephalo-caudal view, the setup is flat, but for the right and left medial lateral oblique, the entire setup can be rotated at an angle about the posterior – anterior axis.

Figure 2. Image splicing

(A–C) are the partial and spliced images from the top probe. (D–F) are the partial and spliced images from the bottom probe. (G) is the fully spliced image. (C, F, G) are normally shown for interpretation to allow for mass detection by shadows and assessment of shadow effects. For comparison with DBT images, (G) is preferable. The green arrows show the location of the cyst in images where it is more difficult to see. Comparing the cyst in (F) and in (E), we see that in (F) it is blurred on the left side and decreased in width relative to that in (E). The blurring and width decrease indicates that the breast moved backwards towards the chest wall in between the POS sweep and the MID sweep. This relatively small error is deemed acceptable, but still demonstrates the difficulty of imaging with multiple shifted sweeps.

Figure 3. Multimodality research workstation display

(A) cephalo-caudal DBT, (B) medial lateral oblique DBT, (C–D) medial lateral oblique US on the combined DBT/US system, shown in two orthogonal views, (E–F) medial lateral oblique US on the dual-sided BLUCI system, also shown in two orthogonal views.

Figure 4. Regions for CNR measurement in decompressed image

(A) is S_{out} , the background volume outside each side of the mass but inside the box circumscribing the mass. (B) is S_{in} , the inner volume of the mass.

Figure 5. Determination of total coverage volume

The coverage volume (A) in the TOP image is the volume between the segmented boundary surface (B) and the pre-segmented skin line (C). Likewise, the BOT coverage volume (D) is the volume between the segmented boundary surface (E) and the pre-segmented skin line (F). After the TOP and BOT coverage volumes are determined, the total coverage volume can be determined as the union of the TOP and BOT coverage volumes (G).

Figure 6. Example Segmentation

The area above the white line of x's is considered to pass the quality criterion and the area below is considered to fail. Such a white line is drawn on several slices, and then these lines are interpolated to form a surface separating the entire breast volume that passes from that which fails. It should be noted that the skin line is not perfectly segmented, but this error will be very small compared to the total volume segmented, and thus can be ignored.

Figure 7. Plot of CNR versus distance from transducer

For each mass, the CNR in the TOP image (downwards-pointing triangles) and the BOT image (upwards-pointing triangles) are plotted as a function of the distance from the corresponding transducer. Points corresponding to simple cysts are triangles that are filled in, while points corresponding to complex cysts and solid masses are triangles that are not filled in. The solid line is a linear fit for the simple cysts, while the dashed line is a linear fit for the complex cysts and solid masses.

A

B

Figure 8. Two examples of masses seen only by one of the probes

(A) is a case where the mass is only in the field of view of the top probe images. This mass corresponds to mass #3 in Table 2. (B) is a case where the mass is in range of both probes but is only visible from the bottom. This mass corresponds to mass #9 in the Table 2. It should be noted that both (A) and (B) are log-compressed decibel images rather than the log-decompressed images that are used for the quantitative CNR measure. The log-compressed images are the images that the radiologists will see, and are shown to provide a qualitative picture of the observed improvement in CNR.

Table 1**Linear fit information**

The slope, y-intercept, and coefficient of determination (R^2) are given for the linear fits to the simple cysts and to the complex cysts and solid masses.

Mass Type	Slope (1/cm)	y-intercept	R^2
Simple cysts	-0.436	2.260	0.499
Complex cysts and solid masses	-0.212	1.514	0.437

Table 2

CNR data for all masses

For each mass, the distance of the mass from each transducer is given for comparison with the CNR of the mass in the TOP and BOT images. Note that for the TOP and BOT, the CNR is given as zero if the mass was out of range of the respective transducer. The smaller of the distance from the top transducer and the distance from the bottom transducer (minimum distance from transducer) is given for comparison with the CNR of the mass in the TOPBOT spliced image. The diameter and classification of the masses are also given. The table is ordered by increasing minimum distance from transducer, and grouped in simple cysts, and simple cysts and solid masses.

Mass No.	Distance from top transducer (mm)	Distance from BOT transducer (mm)	Minimum distance from transducer (mm)	TOP CNR	BOT CNR	TOPBOT CNR	Diameter (mm)	Classification
Simple cysts								
1	19.8	73.8	19.8	1.81	0	1.81	9.0	Simple cyst
2	61.0	21.4	21.4	0	1.12	1.12	10.2	Simple cyst
3	21.4	58.6	21.4	1.09	0	1.09	7.7	Simple cyst
4	68.8	25.8	25.8	0	0.91	0.91	31.0	Simple cyst
5	55.6	27.6	27.6	0	1.27	1.27	11.6	Simple cyst
6	27.6	55.0	27.6	1.25	0	1.25	11.7	Simple cyst
7	29.2	35.4	29.2	0.80	0.77	1.11	20.5	Simple cyst
8	33.2	60.0	33.2	0.81	0	0.81	10.9	Simple cyst
Complex cysts and solid masses								
9	36.4	5.6	5.6	0.19	0.97	0.98	11.7	Complex cyst
10	29.0	13.8	13.8	1.20	1.49	1.65	9.7	Complex cyst
11	44.0	15.2	15.2	0.54	1.22	1.24	11.1	Complex cyst
12	16.4	58.6	16.4	1.28	0	1.28	11.8	Solid mass
13	27.0	33.3	27	1.03	1.03	1.18	11.3	Solid mass

Table 3
Fractional coverage with adequate image quality

The means and standard deviations (σ) of the fractional coverages for the TOP, BOT, and spliced TOPBOT images are given. It is noted that for both the inter-reader and intra-reader differences, the sign of the mean percent difference for the TOP and BOT is opposite that of the TOPBOT. This is possible because the TOPBOT volume covered depends not only on the volumes of the TOP and BOT coverages separately, but also on their overlap, as shown in Figure 7.

Fractional Coverage (25 image sets)	Mean (%)	σ (%)
TOP	59.1	14.5
BOT	58.6	18.3
TOPBOT	89.1	12.2
Inter-reader Difference (25 image sets)	Mean (%)	σ (%)
TOP	1.2	9.4
BOT	3.5	12.0
TOPBOT	-2.4	5.9
Intra-reader Difference (4 image sets)	Mean (%)	σ (%)
TOP	1.4	3.4
BOT	1.6	4.6
TOPBOT	-0.5	0.7