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Receiver operating characteristic analysis for the detection of simulated microcalcifications on mammograms using hardcopy images

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Abstract

The aim of this study was to compare mammography systems based on three different detectors—a conventional screen—film (SF) combination, an a-Si/CsI flat-panel (FP)-based detector, and a charge-coupled device (CCD)-based x-ray phosphor-based detector—for their performance in detecting simulated microcalcifications (MCs). 112–150 μm calcium carbonate grains were used to simulate MCs and were overlapped with a slab phantom of simulated 50% adipose/50% glandular breast tissue-equivalent material referred to as the uniform background. For the tissue structure background, 200–250 μm calcium carbonate grains were used and overlapped with an anthropomorphic breast phantom. All MC phantom images were acquired with and without magnification (1.8X). The hardcopy images were reviewed by five mammographers. A five-point confidence level rating was used to score each detection task. Receiver operating characteristic (ROC) analysis was performed, and the areas under the ROC curves (A_z s) were used to compare the performances of the three mammography systems under various conditions. The results showed that, with a uniform background and contact images, the FP-based system performed significantly better than the SF and the CCD-based systems. For magnified images with a uniform background, the SF and the FP-based systems performed equally well and significantly better than the CCD-based system. With tissue structure background and contact images, the SF system performed significantly better than the FP and the CCD-based systems. With magnified images and a tissue structure background, the SF and the CCD-based systems performed equally well and significantly better than the FP-based system. In the detection of MCs in the fibroglandular and the heterogeneously dense regions, no significant differences were found except that the SF system performed significantly better than the CCD-based system in the fibroglandular regions for the contact images.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Early detection of breast cancer by mammography relies on the detection of suspicious microcalcifications (MCs) and small masses. The detection of MCs requires the contrast-to-noise ratios (CNRs) for the MCs to exceed some threshold value (Barten 1999). The CNR can be improved by improving the contrast of the MCs by optimizing the x-ray spectrum (kVp, target and filter) or by increasing the detected x-ray fluence by using more efficient detectors. Currently, screen–film (SF) mammography is still the primary technique used for screening and diagnostic mammography in the United States. In recent years, various digital detectors (Yaffe and Rowlands 1997, Pisano *et al* 2000) have been developed and commercialized for mammography, including computed radiography (CR) systems (Cowen *et al* 1997, Jennings *et al* 1996, Kimme-Smith *et al* 1989, Liu *et al* 2001, Rong *et al* 2002), charge-coupled device

(CCD)-based systems (Evans *et al* 2002, Liu *et al* 2001, Rong *et al* 2002, Rosen *et al* 1997, Tesic *et al* 1999, Vedantham *et al* 2000b) and amorphous silicon flat-panel (FP)-based systems (Liu *et al* 2001, Mainprize *et al* 1999, Obenauer *et al* 2002, Rong *et al* 2002, Suryanarayanan *et al* 2002, Vedantham *et al* 2000a). These digital detectors use different techniques to improve detective quantum efficiency and to overcome the limitations of conventional SF mammography. The digital mammography systems generally offer image quality comparable to that of SF images along with several advantages, including a wide dynamic range, contrast enhancement, high detection efficiency, flexible image post-processing and analysis, image display on softcopy or hardcopy devices, and easy image management. An additional benefit of digital mammography is the facilitation of advanced imaging techniques and applications, including computer-aided detection and diagnosis (Chan *et al* 1999, Doi *et al* 1997, Jiang *et al* 1999, Zheng *et al* 2002), dual-energy mammography (Asaga *et al* 1987, Johns *et al* 1985, Lemacks *et al* 2002), tomosynthesis (Niklason *et al* 1997, Suryanarayanan *et al* 2000) and cone-beam computed tomography breast imaging (Chen and Ning 2002, Chen *et al* 2005).

Several studies have been conducted to investigate the detection and visualization of calcifications for SF and digital mammography using a variety of approaches. For materials, the American College of Radiology (ACR) and the contrast-detail mammography (CDMAM) phantoms have been widely used to study and compare detection performance (image quality) (Berns *et al* 2002, Funke *et al* 1998, Hermann *et al* 2002, Pisano *et al* 2001, Suryanarayanan *et al* 2002). In addition, calcium carbonate (Rong *et al* 2002), aluminium (Carton *et al* 2004, Cowen *et al* 1997, Desponds *et al* 1991) and aluminium oxide (Sabol *et al* 1993) have been used to simulate calcifications and to investigate the performance of mammography systems in detecting subtle lesions. These investigations also reported the minimum size of detectable MCs for SF and digital mammography systems. For SF mammography, Desponds *et al* (1991) reported a minimum visible size in the range of 170–220 μm and Sabol *et al* (1993) observed a minimum detectable size of 200 μm . Cowen *et al* (1997) found that the minimum detectable size was 130 μm for both SF and CR mammography. Rong *et al* (2002) reported that the minimum visible sizes were 110 μm for SF and FP-based digital mammography and 123 μm for CR and CCD-based mammography. The differences in the findings of these studies may be due to several factors, including the intrinsic uncertainties of mammography systems, additional filtrations, varying breast thicknesses and the differing imaging techniques used. In contrast to phantom studies, several clinical trials comparing SF and digital mammography systems have also been reported (Fischer *et al* 2002, Fischmann *et al* 2005, Grunert *et al* 2001, Lewin *et al* 2001, Lewin *et al* 2002, Obenauer *et al* 2002), however, none of them reported the minimum detectable MC size.

A variety of methods were used to assess the detection and visualization of MCs in these investigations. With the ACR phantom (containing six fibres, five groups of specks and five masses), counting the number or calculating the percentage of detected lesions is the standard method. With the CDMAM phantom (consisting of 16-size and 16-contrast gold discs), four-alternative forced choice (4-AFC) analysis is performed, and the contrast-detail curve is plotted with a predetermined threshold indicating detection accuracy. The most common method for evaluating and comparing imaging technologies against human observer performance is receiver operating characteristic (ROC) analysis, which has become a widely accepted tool in medical imaging over the past two decades. The ROC plot is composed of two fundamental aspects: the true positive fraction (TPF, also known as sensitivity) and the false positive fraction (FPF, also known as 1-specificity). The former indicates the ability to detect lesions and the latter describes the ability to avoid mistaking control cases for lesions. An index of accuracy associated with the ROC curve is A_z , which represents diagnostic accuracy as well as detectability and has become a particularly important metric for evaluating diagnostic

procedures. The ROC curves reflect the effects of both lesions and control cases and display a direct visual comparison on a common set of scales.

We previously conducted a study to evaluate several mammography systems under various conditions by using a wide range of MC size and confidence rating scales (Lai *et al* 2005). The advantage of using the confidence scores is the simplicity with which they reflect the contribution of the MCs. To investigate the effect of MC size on mammograms, we used a wide range of MCs (90–180 μm for uniform background and 160–355 μm for tissue structure background). In the study, it was expected that there would be a plateau for small MCs (whose scores are about 1) and a saturated curve for large MCs (whose scores are about 5) and that there would be a linear shift between the two. It was shown that, with contact imaging, the plateau occurred when the MC size was between 90–125 μm and 160–200 μm or so for the uniform and tissue structure backgrounds, respectively, and 150–180 μm and 250–355 μm for the saturated curve. With magnified imaging, the plateau was located between 90–112 μm and 160–200 μm for the uniform and tissue structure backgrounds, respectively, and 140–180 μm and 224–355 μm for the saturated curve. However, several disadvantages were observed from the curves while using a wide range of MC sizes and confidence scores. First, no control and only MC cases were used in the study, so that the scores reflected only the effects of MCs (compared with the true positive fraction in an ROC curve) and none of the effects of controls (such as the false positive fraction in an ROC curve). Without including the effects of controls, the confidence scores cannot reflect the true performance. Under this circumstance, the confidence scores may not be appropriately used as a metric for detectability (such as A_z under an ROC curve) because the confidence score method may either underestimate or overestimate system performance. Second, the use of a wide range of MC sizes may result in large variations. In evaluating and comparing the detection performances between mammography systems in the previous study, the scores from overall MC sizes were averaged. The analysis showed that the scores were about 1 for the smaller MCs and about 5 for the larger MCs. Thus, neither the smaller nor larger MCs are useful for differentiating detection performance. In addition, while the smaller and the larger MCs were included in computing the average scores, both contributed little impact but large standard deviations. As a result, the use of a wide range of MC sizes may reduce the power in statistical analysis and result in different findings. Therefore, appropriate selection of normal and abnormal cases is very important (Metz 1989, Gur *et al* 1990) when evaluating the detection performance of an imaging system. Based on the findings in our previous study, the appropriate MC range (abnormal cases) can be determined at the linear shift of the curve.

Another important but less understood aspect of mammography is how breast density influences the detection of subtle MCs. It is widely suspected that SF mammography is less sensitive for dense breasts, and several clinical studies have supported that suspicion (Kotre 1998, Mandelson *et al* 2000, van Gils *et al* 1998). The use of digital mammography may help overcome this problem because of wide dynamic range and post-image processing. Boone and Lindfors (2001) have reported the detectability of masses in different breast tissues. However, to our knowledge, a practical and systematic approach for quantifying MC detectability as a function of breast density has not been reported yet.

There are two purposes of this study: one is to evaluate and compare the MC detectability for SF and digital mammography systems by conducting a human observer study and the other is to investigate MC detection in different breast tissues for three mammography systems. To achieve these goals, an appropriate MC size range was selected and overlapped with a uniform and tissue structure backgrounds. An ROC study was then performed to evaluate the diagnostic performances. Also, we undertook this study to determine the minimum detectable MC sizes for these mammography systems by using the A_z value. With tissue structure background,

simulated MC clusters were placed in regions of two different breast densities, fibroglandular and heterogeneously dense, based on the BI-RADS classification system.

2. Materials and methods

2.1. Imaging systems

Images were acquired with a FP-based full-field digital mammography (FFDM) system (SenoGraphe 2000D, General Electric Medical Systems, Milwaukee, WI), a CCD-based small-field digital mammography system (SenoVision, General Electric Medical Systems, Milwaukee, WI) and a conventional SF system (MinR-2000, Eastman Kodak Company, Rochester, NY). The measurements of the modulation transfer function (MTF), noise power spectrum (NPS) and detective quantum efficiency (DQE) of the FP-based and CCD-based systems used in this study were performed in our laboratory previously and reported by Liu *et al* (2001). The specifications are listed in table 1. The three mammographic systems employed identical x-ray tubes, filters, generators and an integrated Bucky grid (31 lines cm^{-1} , 5:1 grid ratio). A dual-track x-ray tube with both molybdenum and rhodium anodes was used. Large (0.3 mm) and small (0.1 mm) focal spots were used for contact and magnification mammography, respectively. The x-ray source-to-image receptor distance was 66 cm, and the source-to-magnification plate distance was 39.7 cm, resulting in a magnification factor of 1.8.

2.2. Simulated microcalcification phantom

Calcium carbonate grains (Computerized Imaging Reference Systems, Norfolk, VA) of various sizes were used to simulate MCs: 112–150 μm for uniform background and 200–250 μm for tissue structure background (table 2). The selection of the MC sizes was based on the result of our previous study (Lai *et al* 2005). Using transparent tape, the MCs from the same size group were attached to $2 \times 2 \text{ cm}^2$ pieces of clear film (developed from unexposed Min-R 2000 film) to form an MC insert. Each MC insert consisted of six MCs, which were arranged in a line shape or in hexagonal shape to simulate malignant and benign MC clusters. The spacing between two adjacent MCs was 2 mm, resulting in a 1 cm long line cluster or a 4 mm wide hexagonal cluster, as shown in figure 1(a). In some clusters (for both shapes), one of the six MCs was removed to create a negative control, as shown in figure 1(b). Because the field-of-view (FOV) for the CCD detector was $6 \times 6 \text{ cm}^2$ for both contact and magnification mammography, the area for placing MC inserts was limited in a $4.2 \times 4.2 \text{ cm}^2$ area to accommodate for all detectors. This area was then divided into four $2 \times 2 \text{ cm}^2$ quadrants. Three MC inserts, each containing MCs from the same size group, were randomly selected and placed in three of the four quadrants. A clear film containing no MCs was placed in the fourth quadrant and served as the control (six negatives), as shown in figure 1(c). Placement of the four inserts was varied to produce 14 phantom images with different patterns. In addition, the line clusters were oriented horizontally, vertically or diagonally with respect to the chest wall. The different shapes and orientations served to vary the MC locations to prevent the readers from randomly selecting and searching six MCs for each quadrant.

2.3. Simulation of two backgrounds

To simulate clinical breast imaging tasks, the simulated MCs were overlapped with either uniform or tissue structure backgrounds. A 5 cm thick slab of 50% adipose and 50% glandular breast tissue equivalent material (Computerized Imaging Reference Systems, Norfolk, VA) and an anthropomorphic breast phantom (RMI 165, Gammex RMI, Middleton, WI) were used to provide a uniform background and a tissue structure background, respectively. The latter was imaged to generate the appearance of a medium density breast with an average attenuation equivalent to that generated by 5 cm of 50% adipose and 50% glandular breast tissue. The MC phantoms were overlaid with the anthropomorphic breast phantom in such a way that three quadrants were located in the scattered fibroglandular regions (25–50% glandular tissue over

the entire region) and one quadrant was located in a heterogeneously dense region (50–75% glandular tissue over the entire region), according to the BI-RADS classification of parenchymal patterns (American College of Radiology 2003).

2.4. Image acquisition and review

During image acquisition, a phantom holder consisting of three layers of developed unexposed films, as shown in figure 2, was used to position and stabilize the MC phantom. On the top layer, three rails were attached and used to align the slab or the anthropomorphic breast phantom for reproducible positioning. In the middle layer, four $2 \times 2 \text{ cm}^2$ cutouts were used to accommodate the MC phantom and keep the MC cluster positions fixed in relation to the overlapping tissue structure when the slab or the anthropomorphic breast phantom was imaged. The phantom holder was centred and placed against the chest wall edge of the supporting plate for contact imaging or the magnification plate for magnification imaging. The Bucky (with grid) was used for all contact images but not for magnification images. The slab or the anthropomorphic breast phantom was placed on top of the phantom holder.

To determine the exposure used, a 50% adipose and 50% glandular 5 cm slab phantom was imaged with a screen–film technique using automatic exposure control (AEC). The contrast option was selected on the AEC with a Mo–Rh target–filter combination at 28 kVp and a prescribed recommendation for using 93 mAs. Because the technique used to acquire the images was operated manually, several consensus images of slab and anthropomorphic breast phantoms were acquired with the SF mammography at the exposure setting at 100 mAs, and they were examined by two radiologists prior to performing the reader study. Fourteen MC phantom patterns were acquired with the SF and the two digital mammography systems. The exposed films were processed on a Kodak X-OMAT Multiloader 300 processor (Eastman Kodak Company, Rochester, NY). The CCD and the FP images were processed with the manufacturers' clinical protocols (such as the settings for window level and width) and printed on $8 \times 10 \text{ inch}^2$ laser films (Scopix LT2B, Agfa Medical Imaging, Wilmington, DE) using a laser film printer (Scopix LR5200, Agfa Medical Imaging). The printer had a pixel size of $43 \mu\text{m}$, which should be adequate for printing the CCD images (pixel size $30 \mu\text{m}$). Regarding the image processing for the digital mammograms (FP and CCD images) and the quality of the laser printer and the laser film used in this study, these features were monitored through the collaborative efforts of our clinical imaging physicists and the manufacturers. To ensure that the image quality used in this study is similar to that in clinical use, several consensus printed digital images were examined by a radiologist to determine that the film density and the contrast were at the proper levels prior to printing.

All hardcopy SF and digital images were displayed on a dedicated motorized mammogram viewer (MS604A, S&S RADX Technology, Houston, TX) in 12 groups of various combinations of imaging systems, magnification modes and background types for each reading session. In order to minimize the bias due to temporal memory, the viewing sequence for each reading session was randomized and the interval between consecutive reading sessions was one week. Five mammographers independently reviewed the images, and they were allowed to use a magnifying glass. Clear films with marks to indicate the possible locations and the shape of the MC cluster for all images (SF, FP and CCD) were provided to the observers so that the observers did not have to search over the entire image area for the MCs while reviewing the images. The readers rated the visibility of each possible MC location using a five-point scale: 1 = definitely not present, 2 = probably not present, 3 = possibly present, 4 = probably present and 5 = definitely present.

2.5. Data analysis

The scores were sorted according to the background type, magnification mode, MC size group, imaging system and reader prior to analysis. A mixed factorial ANOVA of the form suggested by Dorfman (Dorfman *et al* 1992,1998) was used on the raw scores to calculate the differences in MC size and the variability of the mammographers. Statistics were calculated using SPSS, version 11.5 (SPSS, Inc., Chicago, IL). ROC analysis was performed and the areas under the ROC curves (A_z s) were computed for various combinations of background types, magnification modes and imaging systems for individual mammographers as well as for individual MC size groups. The ranges of the A_z for the individual readers along with the mean values are listed in tables 3 (uniform background) and 4 (tissue structure background) for various imaging system–magnification combinations. The A_z s for all readers combined are plotted as a function of the average MC size in figure 4. Additionally, the A_z s for all size groups combined are plotted in figures 5(a) and (b) for individual readers and for all readers combined with uniform and tissue structure backgrounds, respectively. To show the detection of MCs located in different breast compositions, the A_z s for all size groups and all readers combined were plotted for various imaging system–magnification combinations in figure 6. Table 5 lists the A_z s for all size groups and all readers combined for various imaging system–magnification combinations.

A two-tailed paired-sample *t*-test was performed to compute the *p*-values to assess the statistical significance of the differences in the A_z between each pair of imaging systems for various conditions. The *p*-values were also computed to evaluate the performance between the contact and the magnified images for each imaging system. Additionally, the *p*-values were also computed to evaluate the performance differences between imaging systems in the scattered fibroglandular regions and the heterogeneously dense regions. The differences were considered statistically significant when $p < 0.05$.

3. Results

Images of the simulated MCs are shown in figures 3(a) and (b) for the uniform and the tissue structure backgrounds, respectively. In each figure, both contact (1X, top) and magnified (1.8X, bottom) images are shown with all MC clusters marked with a square to indicate their locations. The images in figure 3 were acquired with the SF (left), the FP-based (middle) and the CCD-based (right) systems, in which the analogue SF images were digitized with a laser film scanner operating at 600 dpi. The range of the A_z for individual mammographers and the average A_z s are listed in tables 3 and 4 for the uniform and the tissue structure backgrounds, respectively, for various detector systems and magnification modes for individual MC groups.

3.1. A_z versus MC size

The A_z s for all readers combined are plotted as a function of the average MC size for the six imaging system–magnification combinations in figure 4. The A_z increased with the MC size but with slightly different incremental rates for different conditions. In contact imaging with a uniform background, the increments in A_z from the medium to the large size were statistically significant ($p < 0.05$) for all systems. With magnification added, the increments of A_z were significant ($p < 0.05$) from the small to the medium and from the medium to the large size for all systems. In contact imaging with a tissue structure background, the increment in A_z from the medium to the large size was statistically significant ($p < 0.05$) only for the CCD-based system. With magnification added, no significant increment was found from the small to the medium and from the medium to the large size for any of the systems.

3.2. Comparison of imaging systems

To compare the three systems, the A_z s for all MC size groups combined are plotted for individual readers and for all readers combined with a uniform background in figure 5(a) and

with a tissue structure background in figure 5(b). Table 5 lists the A_z s for various imaging detector–magnification combinations and for all readers combined along with the p -values computed to assess the statistical significance of the differences in the A_z between imaging systems.

3.2.1. Uniform background, contact mode.—The FP-based system received the highest A_z for four of five readers and for all readers combined (0.61 ± 0.03), while the SF (0.55 ± 0.02) and the CCD-based (0.53 ± 0.01) systems were the second and third, respectively, for all readers combined (figure 5(a), table 5). The difference in A_z between the FP-based and the SF or the CCD-based systems was statistically significant ($p < 0.01$). The difference in A_z between the CCD-based and the SF systems was statistically insignificant ($p = 0.3$). These findings indicated that the FP-based system clearly outperformed both the SF and the CCD-based systems.

3.2.2. Uniform background, magnification mode.—The performance of the SF system improved substantially (from 0.55 to 0.76, table 5), but the performances of the FP-based (from 0.61 to 0.76) and the CCD-based (from 0.53 to 0.70) systems improved only moderately. The performance of the SF system approached that of the FP-based system ($p = 0.89$), and the CCD-based system was outperformed by the FP-based and the SF systems ($p < 0.05$ for both).

3.2.3. Tissue structure background, contact mode.—The SF system received the highest A_z in four of five readers and for all readers combined (0.85 ± 0.02 , table 5), while the FP-based system was the second for all readers combined (0.81 ± 0.02) and for three of five readers (figure 5(b)), and the CCD-based system was third for all readers combined (0.80 ± 0.02). The difference in A_z between the SF and the FP-based or the SF and the CCD-based systems was statistically significant ($p < 0.05$). The difference in A_z between the FP-based and the CCD-based systems was not statistically significant ($p = 0.9$).

3.2.4. Tissue structure background, magnification mode.—The SF system had the highest A_z for all readers combined (0.89 ± 0.02 , table 5) and for three of five readers (figure 5(b)). The CCD-based (0.88 ± 0.01) and the FP-based (0.85 ± 0.02) systems were the second and third, respectively, for all readers combined. The performance of the CCD-based system improved substantially (from 0.80 to 0.88), but that of the SF (from 0.85 to 0.89) and the FP-based (from 0.81 to 0.85) systems improved only moderately. The performance of the CCD-based system approached that of the SF system ($p = 0.97$), and the FP-based system was outperformed by both the SF and the CCD-based systems ($p < 0.05$ for both).

3.3. A_z versus breast density

To compare the system performances in the detection of MCs with different breast compositions (densities), figure 6 shows the A_z s for two different background types, fibroglandular and heterogeneously dense, for all readers and all MC size groups combined. In addition, the A_z for each system–magnification combination are listed in table 6. As expected, the A_z s with fibroglandular regions were significantly higher ($p < 0.05$, figure 6 and table 6) than those with heterogeneously dense regions for all cases except for the CCD-based system in magnification imaging, which indicated that detection performance may decrease as a result of denser breast composition. Table 6 lists the p -values for comparison between any two imaging systems.

3.3.1. Fibroglandular region.—With contact imaging, the SF system with the highest A_z (0.89 ± 0.01) performed similarly to the FP-based system (0.87 ± 0.02 , $p = 0.4$) but significantly better than the CCD-based system (0.84 ± 0.02 , $p < 0.05$). No significant difference was found between the FP-based and the CCD-based systems. With magnification imaging, the SF system

still had the highest A_z (0.93 ± 0.02), followed by the CCD-based (0.90 ± 0.01) and the FP-based systems (0.89 ± 0.02). However, the differences in A_z between the three systems were not statistically significant ($p > 0.05$).

3.3.2. Heterogeneously dense region.—With contact imaging, the SF system had the highest A_z (0.80 ± 0.03) while the CCD-based (0.76 ± 0.04) and the FP-based (0.74 ± 0.03) systems were the second and third, respectively. The differences in A_z among the systems were not statistically significant ($p > 0.05$). With magnification imaging, the CCD-based system (0.87 ± 0.04) was superior to the SF (0.84 ± 0.04) and the FP-based (0.79 ± 0.04) systems, but not at a statistically significant level ($p > 0.05$).

4. Discussion

It is inevitable to have variations while performing an observer study. These variations result from many factors, especially inter-reader variability due to human observers. Significant inter-reader variability may result in large deviation and less statistical power. Under this circumstance, the relative performances between mammography systems provide another approach to compare different mammography systems under various imaging situations. For example, figure 5(a) shows the significant inter-reader variability in the A_z values for each combination of mammography systems and magnification modes. Figure 5(a) also shows the relative performances of different mammography systems for individual readers. For contact images, readers 2, 3 and 5 had the same ranking order (FP, SF and CCD) and reader 4's order was different (FP, CCD and SF). There is little difference among readers 2–5, and they all reported that the FP system ranked the highest. For magnified images, readers 2, 3 and 4 had the same ranking order (FP, SF and CCD) and reader 5 had different rank order (SF, FP and CCD). Similarly, they all reported that the CCD system had the lowest ranking order. The results for reader 1 were significantly different from the other observers. However, this difference was not reflected in the averaged A_z values because the inter-reader variations can be reduced by averaging the A_z from all readers. Therefore, the ranking order for all readers combined is consistent with that for most individual readers.

In this study using simulated MCs overlapped with uniform and tissue structure backgrounds, we found that the FP-based system performed either better than (1X) or similarly to (1.8X) the SF and the CCD-based systems with a uniform background but worse with a tissue structure background. This change of performance may be attributed to several possible factors. First, one advantage of digital mammography techniques is that they allow flexible post-processing for optimal display or printing. This feature was used to the greatest advantage with the uniform background as the window/level settings were adjusted to print the largely flat background and MCs. Thus, the latter were printed with higher contrast (narrower window settings). With the tissue structure background, the window/level settings were adjusted to accommodate the large signal range of the entire image. Under this circumstance, the MC images were printed with a lower contrast level (wider window settings). This may have contributed to the poorer performance of the FP-based system with a tissue structure background. Therefore, the detection of MCs with the use of softcopy images from digital mammography systems will likely be improved, but further study is needed.

Another explanation could involve the obscuring effect of the overlapping tissue structure background. With the uniform background, the detection and visualization of the MCs were dictated by the contrast-to-noise ratio (CNR) for the object (not the individual pixel). The higher the CNR values, the better the MCs could be detected and visualized. For quantum-limited imaging situations, the CNR value is approximately proportional to the square root of the number of detected photons within the projected MC image on the detector. Therefore, imaging systems with higher detective quantum efficiency (DQE) lead to better signal-to-noise ratios

(SNRs) and CNRs. The FP-based system employs a more absorptive CsI:Tl scintillator, in contrast to the less absorptive GdSO₄:Tb screens used in the SF and the CCD-based systems. This results in a better x-ray absorption ability and a higher DQE for the FP-based image receptor than for the SF and the CCD-based image receptors, leading to better SNRs and CNRs. Therefore, the FP-based system was expected to and did perform the best with a uniform background. However, with the tissue structure background, the detection of MCs was more difficult because of the increased structural noise from the presence of tissue structures rather than the CNR from the MC itself. As a result, the higher spatial resolution or better modulation transfer function (MTF) of the SF system could play an important role in differentiating MCs from the overlapping tissue structures, resulting in a sharper appearance of the MCs, which may help to improve MC visibility.

Overlapping tissue structures not only affect the relative performances of different imaging systems but also alter the significance of performance improvement due to magnification. The use of magnification has been shown to improve visualization of MCs because of a reduced relative noise level and an increased apparent size at the detector plane (Shaw *et al* 2000). Therefore, magnification imaging is especially useful for enhancing the visibility of MCs and improving diagnostic accuracy. Table 7 lists the ratios of A_z with magnification to those without magnification. The ratios show that there was improvement with all imaging systems and with all MC sizes. However, the improvements varied with the MC size and the background type. With a uniform background, the ratios for all imaging systems ranged from 1.23 to 1.54 for 125–140 and 140–150 μm MCs and from 1.06 to 1.11 for 112–125 μm MCs. The significantly smaller improvement with the 112–125 μm MCs may be because the imaging systems had approached their limit in detecting small calcifications. This is well reflected by the low values of A_z (0.45–0.55 and 0.48–0.67 for contact and magnified images, respectively). Thus, there was some improvement with magnification but the amount of improvement was limited by the difficulty of the detection tasks.

With a tissue structure background, the improvement resulting from magnification was small for all imaging systems, with an improvement ratio ranging from 1.01 to 1.15. However, since the A_z for these detection tasks were quite high, ranging from 0.72 to 0.97 for contact images (table 4), and the MCs were larger than those studied with uniform background, the reason for the low level of improvement cannot be attributed to the limited imaging capability of the systems. Instead, it may be explained by the fact that the detection tasks were limited by the effect of the structural noise rather than the CNR of the object. The CNR can be improved by magnification because an apparently larger object is shown in the image and more x-ray photons fall into the object area on the detector. The structural background, however, is little affected by the magnification process, as both the object (MC) and the overlapping structure background are magnified in a similar way, resulting in essentially the same image pattern seen in the non-magnified image (MCs embedded in the tissue structure background) except that the image is enlarged.

In this study, the SF and the CCD mammography systems performed better than the FP mammography system while tissue structure is present, which shows that the image receptor with smaller pixel size has better detection in subtle MCs. However, this does not imply that detection performance can be improved by reducing the pixel size, which can be explained by the following fact. For a given image receptor, reducing pixel size may improve the MTF (spatial resolution), however, the noise level may also increase. Under this circumstance, it is crucial to draw a decisive conclusion without a complete investigation by measuring physical property or by performing an observer study. However, a detailed investigation of the effect of the pixel size on detecting MCs is beyond the scope of this study.

Although there were several differences between these two studies, including the analysis methods (ROC analysis versus confidence score method) and the ranges of the MC sizes, both studies showed similar results: with a uniform background, the FP-based system performed the best followed by the SF system and the CCD-based system, and with a tissue structure background, the result were just the opposite. This can be explained by the detection of MCs associated with the MC size. Large MCs received a higher score with the confidence level method as well as a higher A_z in the ROC study, which was also observed by Rong *et al* (2002), even though control cases were not included in the confidence level method. In addition, the confidence scores and the A_z were calculated by averaging overall MC sizes, in which the equivalent MC size can be regarded as the mean MC size. The mean MC size is 135 μm and 258 μm for the uniform and the tissue structure backgrounds, respectively, for the confidence level method, and 131 μm and 225 μm for the ROC study. The equivalent MC sizes for the studies were similar. Because the detection of MCs depends on the value of SNR^2 and the same techniques were used for both studies, it was expected that the studies would reveal similar SNRs (135 versus 131 μm and 258 versus 225 μm for the uniform and the tissue structure backgrounds, respectively), which they did.

There are several key observations in this study. First, the minimum size of detectable MCs can be determined by using the A_z value, which is often used as a substitute for detectability for an ROC curve. A threshold must be set, and linear interpolation is applied on the detection curves (in figure 4) to obtain the minimum detectable size. A threshold of 92%, corresponding to a detectability of 2, was selected in this study. Under this assumption, the determination of a minimum detectable MC size can be made for magnification mammography. With a uniform background, the minimum detectable MC sizes are 143 and 145 μm for the SF and the FP-based mammography systems, respectively, and with the tissue structure background, 228 μm for the SF and the CCD-based mammography systems. Because the studies used different approaches, including imaging technique, imaging display and simulated backgrounds, there is no intention to compare our results with those of other investigations.

Another important observation is the differences between the SF system and the two digital mammography systems in detection performances (represented by A_z) in dense breasts. Our study did not show that digital mammography systems improved accuracy (A_z) in dense breasts (which may be because hardcopy was used for the digital images in this study); however, a significantly reduced detection performance in dense breast tissue was observed for all mammography systems. These findings can be quantified by calculating ΔA_z ranging from 0.08 to 0.13 for contact images and from 0.03 to 0.10 for magnified images (table 6). It is assumed that the A_z scale is between 0.5 and 1. This indicates that the reduced performance expressed as a fraction is 16–26% $((0.08/0.5) \times 100\%$ and $(0.13/0.5) \times 100\%)$ and 6–20% $((0.03/0.5) \times 100\%$ and $(0.10/0.5) \times 100\%)$ for contact and magnified images, respectively. This study showed that breast density has a profound influence on the detection of MCs. However, a limited range of simulated MCs and breast densities were studied. Therefore, further study in the detection of simulated MCs with a wide range of sizes and a wide range of breast densities (four categories in BI-RADS) is suggested.

5. Conclusion

The relative performance of mammography systems depends not only on their physical properties (e.g., MTF, DQE), but also on the use of magnification and the overlying structure background. The findings in this study indicate that the FP-based mammography system performed significantly better than the SF and the CCD-based mammography systems for contact imaging with a uniform background. With the use of magnification, the performance of the SF system was substantially improved to about the same level as that of the FP-based system. The CCD-based system was outperformed by both the SF and the FP-based systems.

The SF system performed significantly better than the FP-based and the CCD-based systems for contact images with a tissue structure background. With magnification added, the performance of the CCD-based system was improved significantly to a level similar to that of the SF system. The FP-based system was outperformed by both the SF and the CCD-based systems. Breast density has a profound influence on the detection of MCs for all mammography systems. The SF system performed better than the FP-based and the CCD-based systems with fibroglandular and heterogeneously dense regions; however, no statistical significance was found except that the SF system performed significantly better than the CCD-based system in the fibroglandular regions for the contact images.

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References

- American College of Radiology. Illustrated Breast Imaging Reporting and Data System (BI-RADS). American College of Radiology; Reston, VA: 2003. 4th edn
- Asaga T, Chiyasu S, Mastuda S, Mastuura H, Kato H, Ishida M, Komaki T. Breast imaging: dual-energy projection radiography with digital radiography. *Radiology* 1987;164:869–70. [PubMed: 3303124]
- Barten, PGJ. Contrast Sensitivity of the Human Eye and its Effects on Image Quality. SPIE Optical Engineering Press; Bellingham, WA: 1999.
- Berns EA, Hendrick RE, Cutter GR. Performance comparison of full-field digital mammography to screen-film mammography in clinical practice. *Med. Phys* 2002;29:830–4. [PubMed: 12033579]
- Boone JM, Lindfors KK. The effect of breast density on cancer detection performance in mammography. *J. Women's Imaging* 2001;3:122–8.
- Carton AK, Bosmans H, Vandenbroucke D, Souverijns G, Van Ongeval C, Dragusin O, Marchal G. Quantification of Al-equivalent thickness of just visible microcalcifications in full field digital mammograms. *Med. Phys* 2004;31:2165–76. [PubMed: 15305471]
- Chan HP, Sahiner B, Helvie MA, Petrick N, Roubidoux MA, Wilson TE, Adler DD, Paramagul C, Newman JS, Sanjay-Gopal S. Improvement of radiologists' characterization of mammographic masses by using computer-aided diagnosis: an ROC study. *Radiology* 1999;212:817–27. [PubMed: 10478252]
- Chen B, Ning R. Cone-beam volume CT breast imaging: feasibility study. *Med. Phys* 2002;29:755–70. [PubMed: 12033572]
- Chen L, Shaw CC, Tu S-J, Altunbas M, Wang T, Lai C-J, Liu X. Cone-beam CT breast imaging with a flat panel detector—a simulation study. *Proc. SPIE* 2005;5745:943–51.
- Cowen AR, Launders JH, Jadav M, Brettle DS. Visibility of microcalcifications in computed and screen-film mammography. *Phys. Med. Biol* 1997;42:1533–48. [PubMed: 9279904]
- Desponds L, Depeursinge C, Grecescu M, Hessler C, Samiri A, Valley JF. Image quality index (IQI) for screen-film mammography. *Phys. Med. Biol* 1991;36:19–33. [PubMed: 2006213]
- Doi K, Giger M, Nishikawa RM, Schmidt RA. Computer-aided diagnosis of breast cancer on mammograms. *Breast Cancer* 1997;4:228–33. [PubMed: 11091604]
- Dorfman DD, Berbaum KS, Lenth RV, Chen YF, Donaghy BA. Monte Carlo validation of a multireader method for receiver operating characteristic discrete rating design: factorial experimental design. *Acad. Radiol* 1998;5:591–602. [PubMed: 9750888]
- Dorfman DD, Berbaum KS, Metz CE. Receiver operating characteristic rating analysis: generalization to the population of readers and patients with the jackknife method. *Invest. Radiol* 1992;27:723–31. [PubMed: 1399456]
- Evans DS, Workman A, Payne M. A comparison of the imaging properties of CCD-based devices used for small field digital mammography. *Phys. Med. Biol* 2002;47:117–35. [PubMed: 11814221]

- Fischer U, Baum F, Obenauer S, Luftner-Nagel S, von Heyden D, Vosschenrich R, Grabbe E. Comparative study in patients with microcalcifications: Full-field digital mammography versus screen-film mammography. *Eur. Radiol* 2002;12:2679–83. [PubMed: 12386757]
- Fischmann A, Siegmann KC, Wersebe A, Claussen CD, Muller-Schimpfle M. Comparison of full-field digital mammography and film-screen mammography: image quality and lesion detection. *Br. J. Radiol* 2005;78:312–5. [PubMed: 15774591]
- Funke M, Breiter N, Hermann KP, Oestmann JW, Grabbe E. Storage phosphor direct magnification mammography in comparison with conventional screen-film mammography—A phantom study. *Br. J. Radiol* 1998;71:528–34. [PubMed: 9691898]
- Grunert JH, Barbey M, Berndt G, Borchert B, Farber A, Gmelin E, Ratmann R, Rautmann B. DIMA enlargement mammography in microcalcifications: A prospective study with ROC analysis. *Eur. Radiol* 2001;11:284–91. [PubMed: 11218029]
- Gur D, King JL, Rockette HE, Britton CA, Thaete FL, Hoy RJ. Practical issues of experimental ROC analysis. Selection of controls. *Invest. Radiol* 1990;25:583–6. [PubMed: 2345092]
- Hermann KP, Obenauer S, Funke M, Grabbe EH. Magnification mammography: a comparison of full-field digital mammography and screen-film mammography for the detection of simulated small masses and microcalcifications. *Eur. Radiol* 2002;12:2188–91. [PubMed: 12195468]
- Jenning RJ, Jafroudi H, Gagne RM, Fewell TR, Quinn PW, Steller Artz DE, Vucich JJ, Freedman MT, Mun SK. Storage-phosphor-based digital mammography using a low-dose x-ray system optimized for screen-film mammography. *Proc. SPIE* 1996;2708:202–32.
- Jiang Y, Nishikawa RM, Schmidt RA, Metz CE, Giger ML, Doi K. Improving breast cancer diagnosis with computer-aided diagnosis. *Acad. Radiol* 1999;6:22–33. [PubMed: 9891149]
- Johns PC, Drost DJ, Yaffe MJ, Fenster A. Dual-energy mammography: initial experimental results. *Med. Phys* 1985;12:297–304. [PubMed: 4010634]
- Kimme-Smith CM, Bassett LW, Gold RH, Gormley L. Digital mammography. A comparison of two digitization methods. *Invest. Radiol* 1989;24:869–75. [PubMed: 2807802]
- Kotre CJ. The effect of background structure on the detection of low contrast objects in mammography. *Br. J. Radiol* 1998;71:1162–7. [PubMed: 10434911]
- Lai CJ, Shaw CC, Whitman GJ, Johnston DA, Yang WT, Selinko V, Arribas E, Dogan B, Kappadath SC. Visibility of simulated microcalcifications—a hard copy based comparison of three mammographic systems. *Med. Phys* 2005;32:182–94. [PubMed: 15719969]
- Lemacks MR, Kappadath SC, Shaw CC, Liu X, Whitman GJ. A dual-energy subtraction technique for microcalcification imaging in digital mammography—a signal-to-noise analysis. *Med. Phys* 2002;29:1739–51. [PubMed: 12201421]
- Lewin JM, D'Orsi CJ, Hendrick RE, Moss LJ, Isaacs PK, Karellas A, Cutter GR. Clinical comparison of full-field digital mammography and screen-film mammography for detection of breast cancer. *Am. J. Roentgenol* 2002;179:671–7. [PubMed: 12185042]
- Lewin JM, Hendrick RE, D'Orsi CJ, Isaacs PK, Moss LJ, Karellas A, Sisney GA, Kuni CC, Cutter GR. Comparison of full-field digital mammography with screen-film mammography for cancer detection: results of 4945 paired examinations. *Radiology* 2001;218:873–80. [PubMed: 11230669]
- Liu X, Shaw CC, Rong XJ, Lemacks MR. Comparison of a-Si:H/CsI flat-panel digital imaging systems with CR and CCD based systems - image quality measurements. *Proc. SPIE* 2001;4320:389–98.
- Mainprize JG, Ford NL, Yin S, Tumer T, Yafee MJ. Image quality of a prototype direct conversion detector for digital mammography. *Proc. SPIE* 1999;3659:398–406.
- Mandelson MT, Oestreicher N, Porters PL, White D, Finder CA, Taplin SH, White E. Breast density as a predictor of mammographic detection: comparison of interval- and screen-detected cancers. *J. Natl Cancer Inst* 2000;92:1081–7. [PubMed: 10880551]
- Metz CE. Some practical issues of experimental design and data analysis in radiological ROC studies. *Invest. Radiol* 1989;24:234–45. [PubMed: 2753640]
- Niklason LT, et al. Digital tomosynthesis in breast imaging. *Radiology* 1997;205:399–406. [PubMed: 9356620]
- Obenauer S, Luftner-Nagel S, von Heyden D, Munzel U, Baum F, Grabbe E. Screen film versus full-field digital mammography: image quality, detectability and characterization of lesions. *Eur. Radiol* 2002;12:1697–702. [PubMed: 12111060]

- Pisano ED, Britt GG, Lin Y, Schell MJ, Burns CB, Brown ME. Factors affecting phantom scores at annual mammography facility inspection by the U.S. Food and Drug Administration. *Acad. Radiol* 2001;8:864–70. [PubMed: 11724041]
- Pisano ED, Yaffe MJ, Hemminger BM, Hendrick RE, Niklason LT, Maidment AD, Kimme-Smith CM, Feig SA, Sickles EA, Braeuning MP. Current status of full-field digital mammography. *Acad. Radiol* 2000;7:266–80. [PubMed: 10766101]
- Rong XJ, et al. Microcalcification detectability for four mammographic detectors: flat-panel, CCD, CR, and screen/film. *Med. Phys* 2002;29:2052–61. [PubMed: 12349926]
- Rosen D, Phillips W, Stanton M, Xie J, O'Mara D, Stewart A. A prototype CCD-based detector for whole breast digital mammography. *Proc. SPIE* 1997;3032:52–8.
- Sabol JM, Soutar IC, Plewes DB. Observer performance and dose efficiency of mammographic scanning equalization radiography. *Med. Phys* 1993;20:1517–25. [PubMed: 8289736]
- Shaw CC, Liu X, Lemacks MR, Rong JX, Whitman GJ. Optimization of MTF and DQE in magnification radiography—a theoretical analysis. *Proc. SPIE* 2000;3977:466–75.
- Suryanarayanan S, Karellas A, Vedantham S, Glick SJ, D'Orsi CJ, Baker SP, Webber RL. Comparison of tomosynthesis methods used with digital mammography. *Acad. Radiol* 2000;7:1085–97. [PubMed: 11131053]
- Suryanarayanan S, Karellas A, Vedantham S, Ved H, Baker SP, D'Orsi CJ. Flat-panel digital mammography system: contrast-detail comparison between screen-film radiographs and hard-copy images. *Radiology* 2002;225:801–7. [PubMed: 12461264]
- Tesic MM, Fisher PM, Munier B. Full field digital mammography scanner. *Eur. J. Radiol* 1999;31:2–17. [PubMed: 10477093]
- van Gils CH, Otten JD, Verbeek AL, Hendriks JH, Holland R. Effect of mammographic breast density on breast cancer screening performance: a study in Nijmegen, The Netherlands. *J. Epidemiol. Community Health* 1998;52:267–71. [PubMed: 9616416]
- Vedantham S, et al. Full breast digital mammography with an amorphous silicon-based flat panel detector: physical characteristics of a clinical prototype. *Med. Phys* 2000a;27:558–67. [PubMed: 10757607]
- Vedantham S, Karellas A, Suryanarayanan S, Levis I, Sayag M, Kleehammer R, Heidsieck R, D'Orsi CJ. Mammographic imaging with a small format CCD-based digital cassette: physical characteristics of a clinical system. *Med. Phys* 2000b;27:1832–40. [PubMed: 10984230]
- Yaffe MJ, Rowlands JA. X-ray detectors for digital radiography. *Phys. Med. Biol* 1997;42:1–39. [PubMed: 9015806]
- Zheng B, Shah R, Wallace L, Hakim C, Ganott MA, Gur D. Computer-aided detection in mammography: an assessment of performance on current and prior images. *Acad. Radiol* 2002;9:1245–50. [PubMed: 12449356]

Figure 1.

(a) Simulated calcifications were overlapped with the uniform background (top) and tissue structure background (bottom) and acquired with the FP-based mammography system with contact mode. The microcalcification size was 140–150 and 224–250 μm for the uniform and tissue structure backgrounds, respectively. The matrix size was 200×200 for all images. Six MCs were arranged in a 1 cm line shape (left) or 4 mm wide hexagonal shape (right). (b) Simulated calcifications were overlapped with the uniform background (top) and tissue structure background (bottom) and acquired with the FP-based mammography system with contact mode. The microcalcification size was 140–150 and 224–250 μm for the uniform and tissue structure backgrounds, respectively. The matrix size was 200×200 for all images. One MC was missing (square used as a control) for a 1 cm line shape (left) or 4 mm hexagonal shape (right). (c) Simulated MC clusters were overlapped with the uniform background (left) and tissue structure background (right) and acquired with the FP-based mammography system with contact mode. The MC sizes were 112–150 and 200–250 μm for the uniform and tissue structure backgrounds, respectively. Three MC inserts were randomly arranged in any three of four quadrants and an empty film inserted in the last quadrant, providing six controls.

Figure 2.

The diagram shows the template used to position and sandwich the MC phantom overlapped with a 5 cm 50%/50% slab phantom or an anthropomorphic breast phantom.

Figure 3.

The simulated microcalcification clusters (squared) were overlapped with (a) the uniform background and (b) the tissue structure background, and were acquired with the SF (left), the FP-based (middle) and the CCD-based (right) systems. The MC sizes ranged from 112 to 150 μm for the uniform background and from 200 to 250 μm for the tissue structure background. The matrix sizes for contact (top) and magnified (bottom) images are 1.5 K by 1.5 K, 700 by 700 and 2 K by 2 K for the SF, the FP-based and the CCD-based images, respectively.

Figure 4.

The A_z s for all readers combined are plotted as a function of the average simulated MC size for various imaging system—magnification combinations. The size ranges of 112–150 and 200–250 μm are for the uniform and tissue structure backgrounds, respectively.

Figure 5.

The A_z s for various imaging system—magnification combinations for each individual reader and all readers combined with (a) the uniform background and (b) the tissue structure background. The A_z s were computed for all MC sizes combined.

Figure 6.

The A_z s for different imaging system—magnification combinations for all readers and all MC sizes combined with fibroglandular and heterogeneously dense regions.

Table 1.
Technical characteristics of the three mammographic systems.

Detector type	SF	FP	CCD
Model	Min R-2000	SenoGrappe 2000D	SenoVision
Manufacturer	Kodak	GEMS	GEMS
X-ray converter	GdO ₄ S:Tb	CsI(Tl)	GdO ₄ S:Tb
FOV ($m = 1$)	$18 \times 24 \text{ cm}^2$	$19 \times 23 \text{ cm}^2$	$6 \times 6 \text{ cm}^2$
FOV ($m = 1.8$)	$13 \times 18 \text{ cm}^2$	$14 \times 19 \text{ cm}^2$	$4 \times 4 \text{ cm}^2$
Detector matrix (pixel)		1914×2294	2048×2048
Pixel size (μm)		100	30
Dynamic range		$>10^4:1$	$>4000:1$
Data depth (bits)		14 (linear)	12 (linear)

Table 2.

The size ranges and the average sizes of the simulated microcalcification groups in this study.

MC group	Uniform background	Average size (μm)	Tissue structure background	Average size (μm)
	Size range (μm)		Size range (μm)	
Small	112-125	119	200-212	206
Medium	125-140	133	212-224	218
Large	140-150	145	224-250	237

Table 3. The mean values along with the ranges of the A_z from different readers for uniform background.

MC size group (μm)	Contact images (1X)		Magnified images (1.8X)		CCD based	FP based	CCD-based
	SF		SF				
112-125	Mean	0.486		0.541	0.493	0.567	0.521
	Range	0.45-0.52		0.48-0.63	0.46-0.52	0.48-0.67	0.48-0.57
125-140	Mean	0.526		0.809	0.509	0.801	0.681
	Range	0.47-0.59		0.71-0.93	0.46-0.56	0.67-0.90	0.61-0.83
140-150	Mean	0.633		0.941	0.598	0.919	0.893
	Range	0.49-0.78		0.91-0.98	0.55-0.65	0.84-0.95	0.81-0.98

X = magnification factor.

Table 4. The mean values along with the ranges of the A_z from different readers for tissue structure background.

MC size group (μm)	Contact images (1X)		Magnified images (1.8X)		CCD based	FP based	CCD-based
	SF		SF				
200-212	Mean						
	Range						
212-224	Mean						
	Range						
224-250	Mean						
	Range						

X = magnification factor.

The A_z s for all readers and all MC sizes combined along with the p -values used to compare any two imaging systems in detecting simulated MCs overlapped with two different background types.

Table 5.

	Uniform background		p -value 1X versus 1.8X		Tissue structure background		p -value 1X versus 1.8X
	1X	1.8X			1X	1.8X	
A_z							
	SF	0.549	0.764	<0.01	0.852	0.885	0.13
	FP	0.613	0.762	<0.01	0.807	0.848	0.05
	CCD	0.533	0.697	<0.01	0.804	0.884	<0.01
p -value	SF versus FP	<0.01	0.89		0.02	0.04	
	SF versus CCD	0.30	<0.01		0.02	0.97	
	FP versus CCD	<0.01	0.01		0.90	0.02	
X = magnification factor.							

The A_z s for all readers and all size groups combined along with the p -values by comparing any two imaging systems in detecting simulated MCs placed in two different breast background densities of the anthropomorphic breast phantom.

Table 6.

	Contact images (1X)		p -value Fibro versus Hetero	Magnified images (1.8X)		p -value Fibro versus Hetero
	Fibro	Hetero		Fibro	Hetero	
A_z	SF	0.893	0.01	0.925	0.836	0.02
	FP	0.866		0.895	0.793	
p -value	CCD	0.842	<0.01	0.899	0.866	0.23
	SF versus FP	0.40		0.32	0.63	
	SF versus CCD	0.04		0.43	0.81	
	FP versus CCD	0.49		0.98	0.28	

Fibro = fibroglandular regions.
Hetero = heterogeneously dense regions.

The ratios of the A_2S (magnified versus contact) due to the use of magnification for various imaging systems with uniform and tissue structure backgrounds.

Table 7.

	Uniform background		Tissue structure background		
	112-125	125-140	140-150	200-212	212-224
SF	1.11	1.54	1.48	1.05	1.01
FP	1.10	1.38	1.23	1.03	1.06
CCD	1.06	1.34	1.49	1.10	1.15