Comparison of slot scanning digital mammography system with full-field digital mammography system

Chao-Jen Lai,^{a)} Chris C. Shaw, William Geiser, and Lingyun Chen
*Department of Imaging Physics, The University of Texas M. D. Anderson Cancer Center,
Houston, Texas 77030-4009*

Elsa Arribas, Tanya Stephens, Paul L. Davis, Geetha P. Ayyar, Basak E. Dogan,
Victoria A. Nguyen, Gary J. Whitman, and Wei T. Yang
*Department of Diagnostic Radiology, The University of Texas M. D. Anderson Cancer Center,
Houston, Texas 77030-4009*

(Received 5 July 2007; revised 25 March 2008; accepted for publication 4 April 2008;
published 16 May 2008)

The purpose of this study was to evaluate and compare microcalcification detectability of two commercial full-field digital mammography (DM) systems. The first unit was a flat panel based DM system (FFDM) which employed an anti-scatter grid method to reject scatter, and the second unit was a charge-coupled device-based DM system (SSDM) which used scanning slot imaging geometry to reduce scatter radiation. Both systems have comparable scatter-to-primary ratios. In this study, 125–160 and 200–250 μm calcium carbonate grains were used to simulate microcalcifications and imaged by both DM systems. The calcium carbonate grains were overlapped with a 5-cm-thick 50% adipose/50% glandular simulated breast tissue slab and an anthropomorphic breast phantom (RMI 165, Gammex) for imaging at two different mean glandular dose levels: 0.87 and 1.74 mGy. A reading study was conducted with seven board certified mammographers with images displayed on review workstations. A five-point confidence level rating was used to score each detection task. Receiver operating characteristic (ROC) analysis was performed and the area under the ROC curve (A_z) was used to quantify and compare the performances of these two systems. The results showed that with the simulated breast tissue slab (uniform background), the SSDM system resulted in higher A_z 's than the FFDM system at both MGD levels with the difference statistically significant at 0.87 mGy only. With the anthropomorphic breast phantom (tissue structure background), the SSDM system performed better than the FFDM system at 0.87 mGy but worse at 1.74 mGy. However, the differences were not found to be statistically significant. © 2008 American Association of Physicists in Medicine. [DOI: 10.1118/1.2919768]

Key words: digital mammography, microcalcifications, slot scanning, mean glandular dose, ROC

I. INTRODUCTION

Breast cancer is a major health concern for women in North America.¹ According to the American Cancer Society, breast cancer is the second leading cause of cancer death in women today, and more than 170 000 U.S. women will develop breast cancer in 2007.² Women have a gradually increasing risk of developing breast cancer during their lifetime, especially for those age 50 years and older. Thus, early detection and appropriate treatment of breast cancer can improve the chance of survival. Early detection of breast cancer relies on the detection of suspicious microcalcifications (MCs) and small masses. Currently, x-ray mammography has been an important tool in screening for breast cancer, leading to a reduced mortality rate. Conventional screen/film mammography (SFM) provides good image quality with proper exposures, however, there are several limitations for SFM, including a narrow dynamic range, image artifacts, and image display and management issues. For the past decade, various dedicated solid-state imaging devices have been developed and applied to breast imaging, including charge-coupled devices (CCDs)^{3–6} and amorphous silicon flat panel detectors.^{6–10} The digital detectors based on these technolo-

gies provide better detective quantum efficiency than screen-film or photostimulable storage phosphor plates (or computed radiography).^{5,11–14} The advantages of digital mammography (DM) systems based on these digital detectors include contrast enhancement, capabilities for image postprocessing and analysis, flexible image display, and convenient image management. Despite these advantages and improved image quality for DM, one major problem intrinsic to both DM and SFM systems still remains: Scattered radiation, which reduces the signal-to-noise ratio (SNR) and subject contrast in the image and, thus, degrades image quality and image performance.

The amount of scattered radiation can be quantified by the scatter-to-primary ratio (SPR). In SFM, the SPR can range from 0.25 to 1.2, depending on the x-ray beam spectrum, the field size, the breast composition, and the breast thickness.^{15–17} Several different technologies have been developed to reduce scattered radiation incident on the image receptor, resulting in better image contrast. Currently, the most widely used method for reducing scattered radiation in mammography is the anti-scatter grid. Most current mammography systems use lead septa, fiber interspaced, carbon

fiber cover linear grids with a grid ratio of 3.5-5:1 (named Smit-Röntgen grids), and the SPR can be reduced to between 0.1 and 0.3 for 5-cm-thick 50% adipose/50% glandular breasts.¹⁶⁻¹⁹ Recently, high transmission cellular (HTC) grids made of copper septa, air interspaced with 3.8 grid ratio, have been developed and applied to both SFM and DM systems. It has been reported that HTC grids have a higher contrast improvement factor (CIF) than 5:1 ratio linear grids in the range of 4%–18%.²⁰ However, the use of anti-scatter grids not only reduces scattered radiation but also attenuates the primary x-rays. In order to compensate for the loss of primary x-rays and to maintain the optical density on the H&D curve to obtain an optimal image quality for SFM, it is necessary to use higher exposures, resulting in an increased mean glandular dose. For DM, with its wide exposure latitude, the cost for using a grid is the reduction of SNR due to the loss of primary x-rays.

An alternative to anti-scatter grids for scatter rejection is the use of scanning slot imaging geometry. This method uses a collimated fan beam of x-rays and an array detector where the width of the fan beam matches the area of the detector. As the slot beam and array detector move across the breast, only a portion of the breast is imaged at one time. The width of a slot beam typically ranges from 4 to 12 mm, and the scanning time across the whole breast region is about 4–6 s. The use of scanning slot imaging geometry allows scatter to be rejected with little loss of primary x-rays. In addition, no septa or interspace materials are used in this method to absorb primary x-rays, so that scanning slot imaging techniques also provide the potential to improve dose efficiency. One may expect to reduce more scattered radiation while using a narrower slot width, but the drawback is an increase in the scanning time and the x-ray tube heat loading. On the other hand, increasing the slot width makes scanning time shorter and the x-ray tube usage more efficient but with more scatter. Therefore, the dimension of the slot width directly affects the SPR values.^{16,18,21} Several studies have reported that slot scanning imaging techniques can reduce the SPR to as low as 0.12–0.3 for 5-cm-thick 50% adipose/50% glandular breasts with a 10 mm slot width.^{16,21,22} This indicates that both scatter reduction methods, anti-scatter grids (3.5–5 grid ratio) and scanning slot imaging geometry (10 mm slot width), are able to achieve comparable SPR values.

Two commercial DM systems are available at our institution. One is an *a*-Si:H/CsI(Tl) FP-based full field DM system, which uses an anti-scatter grid to reduce scatter, and the other is a CCD/scintillator-based slot scanning DM system, which employs a scanning slot apparatus for scatter rejection. In this work, an observer performance study was performed to compare these two DM systems in the detection and visualization of simulated MCs. MCs of various sizes were overlapped with two types of backgrounds and imaged at two different mean glandular doses to investigate variations in detection performance with different MC sizes and different dose levels. Receiver operating characteristic (ROC) analysis was performed and the areas under the

curves (A_z s) were computed to evaluate and compare the detection performances between the two DM systems.

II. MATERIALS AND METHODS

II.A. Imaging systems

All image acquisitions in this study were obtained with a FP-based full-field digital mammography (FFDM) system (Senographe 2000D, General Electric Medical Systems, Milwaukee, WI) and a CCD-based slot scanning DM (SSDM) system (SenoScan, Fischer Imaging, Denver, CO). The specifications for these two mammography systems are listed in Table I.

The FFDM system investigated in this study employed a flat panel detector, consisting of a layer of a thallium-activated cesium iodide [CsI(Tl)] scintillator deposited on an array of 1914×2294 $100 \times 100 \mu\text{m}^2$ detector elements. Each detector element contained a photodiode and a thin film transistor (TFT) switch for light detection and signal readout, respectively. The charge signals accumulated during the exposure were individually read out and digitized into 14 bit data with a dynamic range of $>10\,000:1$. The detector was mounted on a Senographe DMR mammographic x-ray unit (General Electric Medical Systems) with an integrated Bucky (31 lines/cm, 5:1 grid ratio). X-rays were generated with a dual-track anode (molybdenum and rhodium), filtered by 0.03 mm molybdenum or by 0.025 mm rhodium for various breast thicknesses and compositions. 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 (SID) was 66 cm. The physical characteristics and performance features of this DM system have been investigated and reported.^{8,23}

The SSDM system investigated in this study used a CsI(Tl) scintillator coupled to an array of rectangular CCDs, operated in the time-delay integration (TDI) mode, through fiberoptic tapers. The CCD array converted light from the scintillator into charge signals, which were shifted along each row in alignment with the scanning direction to allow the exposure signals to be integrated on each detector element. The image signals were read out as 12 bit data with a dynamic range of $>4000:1$. The dimensions of the detector were approximately $22 \times 1 \text{ cm}^2$. The x-ray beam was collimated into a narrow slot to match the detector size. The detector elements had a pitch of $27 \mu\text{m}$ but when sampled were binned to form $54 \times 54 \mu\text{m}$ pixels, resulting in an image matrix size of 4096×5625 pixel for a field of view of $22 \times 29 \text{ cm}^2$, which can image all different breast sizes. The system could also be operated in high resolution mode in which the data are read out with a $27 \mu\text{m}$ pitch and same image matrix size but with a reduced field of view (FOV) of $11 \times 15 \text{ cm}^2$. A tungsten target was used with a nominal focal spot size of 0.3 mm and filtration by 0.5 mm of Al. The SID was 61.8 cm. The effective exposure time (actual exposure time to each point in the image) and the total scanning time were about 0.2 and 6 s, respectively. The physical characteristics of the SSDM system have been investigated by several studies.^{3,24,25}

TABLE I. Physical characteristics and specifications of two digital mammography systems.

Mammography system	FP detector-full field	CCD detector-slot scanning
Model	SenoGraphe 2000D	SenoScan
X-ray converter ^a	CsI(Tl) phosphor	CsI(Tl) phosphor, 150 μm
Pixel size (μm)	100	54
Detector matrix (pixel)	1914 $\times$ 2294	4095 $\times$ 5625
Target-filter	Mo-Rh	W-Al
Focal spot size (mm)	0.3	0.3
SID (cm)	66	61.8
FOV area (cm^2)	19 $\times$ 23	22 $\times$ 29
Bucky or Grid	31 lines/cm, 5:1 grid ratio	N/A
Slot dimension (cm^2)	N/A	22 $\times$ 1
Nyquist frequency (cycles/mm)	5	18.5
Modulation transfer function ^{b, c}	MTF(2)=0.78	MTF(2)=0.82
	MTF(5)=0.3	MTF(5)=0.4
		MTF(9) > 0.1
Detective quantum efficiency ^{b, c}	DQE(0)=0.45	DQE(0) > 0.45
	DQE(2)=0.44	DQE(2)=0.40
	DQE(5)=0.20	DQE(5)=0.31
		DQE(9)=0.13
Data depth (bits)	14	12

^aReference 3.^bReference 21.^cReference 22.

II.B. Simulated background and microcalcification phantoms

Two types of backgrounds were investigated in this study, uniform and tissue structure backgrounds. A 5-cm-thick slab of 50% adipose/50% glandular breast tissue equivalent material (Computerized Imaging Reference Systems, Norfolk, VA), which mimics the attenuation properties of average density compressed breasts,²⁶ was used to simulate a uniform background. The dimension of this uniform phantom was $12.5^W \times 10^D$ cm. In addition, an anthropomorphic breast phantom (RMI 165, Gammex RMI, Middleton, WI) was used to simulate a tissue structure background.²⁷ The breast phantom was composed of a coarse anatomic feature layer, an intermediate layer, and a mercury-intensified film layer to generate the appearance of a normal breast whose average attenuation is comparable to 5-cm-thick 50% adipose/50% glandular compressed breasts. For simulating microcalcifications, three size groups of calcium carbonate grains (Computerized Imaging Reference Systems) were used: 125–160 μm for the uniform background and 200–250 μm for tissue structure background (Table II). The method used to make MC inserts and generate MC phantoms has been

reported in another article.²⁸ Six MCs from the same size group were attached to 2×2 cm^2 pieces of clear film to form a line-shape or a hexagon-shape MC insert [Fig. 1(a)]. However, one of the six MCs was removed to create a negative as the control for some MC inserts [Fig. 1(b)]. Three MC inserts from different MC size groups and an empty insert containing no MCs were randomly positioned to construct a 2×2 MC phantom. The MC phantom was composed of a mixture of line- and hexagon-shape MC inserts. The line-shape MC inserts could be positioned in parallel, diagonal, or perpendicular orientation with respect to the chest wall. The MC phantom was then overlapped with the uniform background [Fig. 2(a)] and the tissue structure background [Fig. 2(b)] for image acquisition. Fourteen different patterns of the MC phantoms were generated. These arrangements were designed to make it difficult for the readers to memorize the MC patterns when analyzing the images to evaluate the presence of MCs.

II.C. Image acquisition and display

In order to provide a comparison to screen-film mammography (SFM) as close a technique as possible was used on

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

MC group	Uniform background		Tissue structure background	
	Size range (μm)	Average size (μm)	Size range (μm)	Average size (μm)
Small	125–140	133	200–212	206
Medium	140–150	145	212–224	218
Large	150–160	155	224–250	237

FIG. 1. (a) Simulated calcifications were overlapped with the uniform background (top) and the tissue structure background (bottom) and acquired with the FP-based mammography system at 1.74 mGy. Six locations each containing MC were circled. (b) Simulated calcifications were overlapped with the uniform background (top) and the tissue structure background (bottom) and acquired with the FP-based mammography system at 1.74 mGy. One location without MC was squared and used as a control.

the SFM unit, the technique for the FFDM was obtained based on particular settings on the FFDM unit. A predetermined exposure was performed for a 5-cm-thick 50% adipose/50% glandular equivalent breast tissue with a SFM system (Senographe DMR, GE Medical Systems) using the system automatic exposure control (AEC). The contrast (CNT) option was selected on the AEC with a Mo–Rh target-filter combination at 28 kVp and the phototimed exposure was 93 mAs. Because the predetermined exposure setting of 93 mAs could not be obtained manually on the console while the AEC was off and it was desired that all of the images in this study would be acquired with the same imaging technique, an exposure setting of 100 mAs was selected

FIG. 2. Simulated MC clusters were overlapped with the uniform background (top) and the tissue structure background (bottom) and acquired with the FP-based mammography system at 1.74 mGy.

and an optical density of 1.5 on the film was obtained. The use of the SFM system to determine the imaging technique was because the FFDM and the SFM systems employ the same model x-ray tube, generator, filter, focal spot, the SID, and an integrated Bucky grid. Using this exposure setting, the entrance skin exposures (ESE) were then measured for the FFDM system (HVL=0.41 mm Al at 28 kVp, Mo–Rh combination) while the phantoms were removed. The mean glandular dose (MGD) can then be estimated from the following equation:

$$D = D_{gN} \times E, \quad (1)$$

where D is the MGD expressed in milligrays (mGy), E is the ESE expressed in roentgens (R), and D_{gN} is the normalized dose conversion factor in mGy/R resulting from an incident exposure in air of 1 R. D_{gN} is a function of breast composition, breast thickness, x-ray beam quality (tube potential and beam HVL), and target-filter combination. Using the D_{gN} values reported for 50% glandular and 50% adipose tissue by Wu *et al.*²⁹ the MGD was estimated to be 1.74 mGy. For the SSDM system, the x-ray tube potential was determined by selecting the AEC mode and 50% glandular-50% adipose tissue while the compressed paddle was positioned at 5 cm above the supporting plate. The tube voltage was found to be 31 kVp. Higher kVp is generally used for the SSDM system compared to the FFDM system due to limitations by heat loading and x-ray transmission. The beam quality and the ESE were measured (HVL=0.47 mm Al). Using the D_{gN} value reported by the NCRP³⁰ for a W-Al target-filter combination, the current setting (mA) was then calculated to obtain the MGD equal (or close) to 1.74 mGy. It was found that the current setting was 180 mA, which means that the

exposure was 1080 mAs or the effective exposure was 36 mAs.

Because one of the advantages of employing scanning slot imaging techniques is dose efficiency, low dose (compared to the normal MGD) imaging was performed with both DM systems to investigate the detection difference between these two DM systems. Based on our previous study,³¹ 0.87 mGy was selected for the low dose. It was found that the techniques used to acquire phantom images at 0.87 mGy were 50 mAs for the FFDM system (at 28 kVp and Mo–Rh combination) and 540 mAs (where the current setting was 90 mA) for the SSDM system (at 31 kVp and W–Al combination).

The MC phantoms overlapped with the uniform and the tissue structure backgrounds were then imaged with both DM systems at the two MGD levels. The raw images were processed with the standard processing algorithms prior to performing the observer study. These algorithms were supplied by the manufacturers and used in clinical work at that time (in 2004). The processed images acquired with the FFDM system were displayed on a General Electric review workstation (RWS) and those acquired with the SSDM system were displayed on a Fischer RWS. Image displays for both systems were based on standard reviewing algorithms provided by the manufacturers. Each RWS was equipped with two high-resolution ($2\text{ K} \times 2.5\text{ K}$) CRT monitors (SMM 21200P made by Siemens for the GE RWS and MGD 521 made by Barco for the Fischer RWS). Calibration procedures (Dome Luminance Calibration v1.7.0 and Dome Calibration TQA v2.0.3.3 for the FFDM and the SSDM systems, respectively) supplied by the manufacturers were performed for the monitors based on Barten curves with the same minimum and maximum luminance levels prior to image display. A photometer (Tektronix J18 with J1811 illuminance head) was used to measure the background illuminance, which was less than 5 lux for both reading rooms.

II.D. Image evaluation and data analysis

Seven mammographers reviewed the softcopy images independently. The mammographers were allowed to use several image manipulation functions, including window/level adjustment, black/white inversion, and zooming. The readings were performed in four separate reading sessions for each observer. The display sequences for each reading session were randomized. The readers rated the visibility of the MC at each possible location and assigned one of the following scores: 1=definitely not present, 2=probably not present, 3=possibly present, 4=probably present, and 5=definitely present. No constraints on time or viewing distance were imposed.

The scores were analyzed with receiver operating characteristic (ROC) methodology using the DBM MRMC program (2.1 Beta) of Berbaum *et al.*^{32–36} The detection performances were measured by computing the area under the ROC curve (A_z). The A_z 's were averaged over all observers and plotted as a function of the MGD level. The statistical significance of the differences in the A_z 's between the two

Fig. 3. (a) The ROC curves for all readers and all MC sizes combined for various imaging system-dose level combinations for the uniform background. (b) The ROC curves for all readers and all MC sizes combined for various imaging system-dose level combinations for the tissue structure background.

DM systems was computed by the two-tailed p values and paired t -test from the DBM MRMC program. The p values were also computed to evaluate the performance difference between the two MGD levels for each mammography system. The differences were considered statistically significant when $p < 0.05$.

III. RESULTS

The ROC curves computed for all observers are shown in Figs. 3(a) and 3(b) for the uniform and the tissue structure backgrounds, respectively. Table III lists the A_z 's for various imaging system-MGD combinations along with the p values for both background types.

TABLE III. The A_z 's for all readers and all MC sizes combined along with the p -values used to compare the two imaging systems in detecting simulated MCs.

		Uniform background			Tissue structure background		
		0.87 mGy	1.74 mGy	p value 0.87 vs 1.74	0.87 mGy	1.74 mGy	p value 0.87 vs 1.74
A_z	FP-full field	0.684 (0.03)	0.803 (0.03)	<0.01	0.819 (0.022)	0.862 (0.021)	0.12
	CCD-slot scanning	0.753 (0.03)	0.836 (0.03)	<0.01	0.843 (0.02)	0.853 (0.015)	0.71
	p value	0.014	0.21		0.38	0.73	

III.A. Comparison of the two dose levels

With the uniform background, the A_z 's increased statistically significantly ($p < 0.05$) with the MGD increasing from 0.87 to 1.74 mGy for both DM systems. With the tissue structure background, the A_z 's also increased with the MGD but not in a statistically significant way ($p = 0.12$ and 0.71 for the FFDM and the SSDM systems, respectively).

III.B. Comparison of the two digital mammography systems

With the uniform background, the SSDM system resulted in higher A_z 's than the FFDM system at both MGD levels (0.753 versus 0.684 and 0.836 versus 0.803 at 0.87 mGy and 1.74 mGy, respectively). However, the differences in the A_z 's between the two systems were statistically significant at 0.87 mGy only. With the tissue structure background, the SSDM system performed slightly better than the FFDM system at 0.87 mGy (0.843 versus 0.819), and slightly worse at 1.74 mGy (0.853 versus 0.862). However, no statistical significance was found for the differences in the A_z 's. Furthermore, a cross comparison between the images acquired with the FFDM at 1.74 mGy and images acquired with the SSDM at 0.87 mGy was made and it showed that the former performed significantly better ($p = 0.031$) than the latter.

III.C. A_z versus MC size

The A_z 's computed for all readers were plotted as a function of the average MC size for the four system-MGD combinations in Fig. 4. With the uniform background, the A_z 's increased with the MC size and the increments were all statistically significant ($p < 0.05$) for all combinations and sizes except for the FFDM with both dose levels from the small to the medium size. With the tissue structure background, the A_z 's increased from the small size to the medium size but not from the medium size to the large size. No statistical significances were found for all scenarios except for the CCD-based SSDM with 1.74 mGy while the MC size increased from the small to the medium size.

IV. DISCUSSION

Important points made by our study are: First, the results with the uniform background showed that the SSDM system resulted in higher A_z 's than the FFDM system at both dose

levels although the difference was not statistically significant at the higher dose level. The results also showed that the performance of both systems improved with increased dose level. This is expected because the scanning slot imaging technique can achieve scatter rejection effectively with little attenuation of the primary x-ray beam, thus resulting in higher contrast signal-to-noise ratios (CNRs). This has been demonstrated with theoretical derivations and imaging experiments.³⁷

Second, with the tissue structure background, the SSDM system resulted in higher A_z only at the lower dose level with a smaller difference. At the higher dose level, the two systems resulted in nearly identical A_z 's. The performance differences at both dose levels were shown to be statistically insignificant. With the dose increase, the A_z for the FFDM system improved with a small difference which was shown to be statistically insignificant. For the SSDM system, the A_z improved with a statistically insignificant difference which was even smaller. This result was related to the fact that the uniform background was replaced with the tissue structure background which may be considered as additional noise, sometimes referred to as anatomical noise. The effect of this noise is well recognized but has not been well understood, despite previous research efforts. Both the scanning slot im-

FIG. 4. The A_z 's for all readers combined plotted as a function of the average simulated MC size for various imaging system-dose level combinations.

aging technique and the anti-scatter grid method were designed to improve the CNRs in the image. With the improvement in the image CNRs, both the MCs and the tissue structures become clearer to visualize and the effects of the anatomical noise remained the same. Thus, with the presence of the tissue structures, the detection and the visualization of the MCs were no longer sole a function of the CNRs as the tissue structures may also be obscuring the MCs. At the lower dose level, both the CNRs and the anatomical noise may have a combined effect on the MC visibility. At the higher dose level, the CNRs would be improved bringing more MCs to the visible level if the background is uniform. However, the anatomical noise now becomes the dominating factor that limits the visibility of these MCs. This may explain why the SSDM system resulted in a higher A_z at the low dose level but a similar A_z at the higher dose level.

Third, one would expect that the A_z increases with MC size increases. However, it was not observed with the tissue background. This may be due to the effects of the background texture which may be more dominating for certain ranges of frequencies or object sizes. As a result, the statistical fluctuation may have caused the unexpected observations. We have computed the p -value for variation of the A_z 's from small to medium size and from medium to large size. The results show that the only significant variation was for the SSDM to increase from small to medium size.

Since the two systems compared in this study employ scanning slot imaging and anti-scatter grid techniques for scatter rejection, one might expect to see the true differences between the scanning slot imaging technique and the anti-scatter grid technique. However, due to the differences in the detectors and the x-ray sources used, the results from this comparison study combined the differences between the two scatter rejection techniques and those between the detectors and the x-ray sources used. The FFDM and the SSDM systems investigated in this study employ the same type of scintillators (CsI) but different light detectors and different signal readout systems. Table I shows that the DQEs at zero frequency do not differ significantly between these two DM systems. This indicates that the x-ray absorption characteristics of the two systems are similar but the readout noise levels may be somewhat different. In addition, no significant difference in the DQEs was found between these two DM systems at spatial frequency <5 cycles/mm. This result implies that the effect of the system noise can be ignored because the exposure level is always on the high side in mammography. Another observation may also indicate the insignificance of the higher system noise of the CCD-based detector in mammography is that in Fig. 4, the SSDM seemed to consistently perform better than the FFDM at the low dose level but equally or more poorly at the high dose level.

The major differences between the two imaging systems may lie with the x-ray sources used. The FFDM system employs a molybdenum (Mo) target filtered by a rhodium (Rh) filter. The SSDM system, probably for the sake of achieving high heat loading, employs an x-ray tube with a tungsten

target and an aluminum filter. The MC contrast for the SSDM image is expected to be lower compared to the FFDM images due to the target and higher kVp. On the other hand, one of the major advantages of DM systems is soft-copy image display with processing tools for viewing. Therefore, the image quality for the SSDM and the FFDM images can be optimized individually. This indicates that the MC contrast may not be a considerable concern. Instead, MC CNR is more relevant relation to the DM performance. Thus, the differences shown by our results with the uniform background may be lower than the true differences between the scanning slot imaging technique and the anti-scatter grid technique.

Another factor that may affect our comparison results is the pixel size. Although both systems employ CsI scintillators and are expected to have similar predigitization modulation transfer functions (MTFs), the smaller pixel size with the SSDM system ($54\ \mu\text{m}$ versus $100\ \mu\text{m}$) helps achieve a higher spatial resolution with its smaller aperture function and smaller sampling distance. This can be confirmed in the MTF values (Table I) and observed in the MC detection performance with the uniform background at low dose for the small and the medium sizes for the SSDM (Fig. 4). However, the higher spatial resolution should have made the SSDM system perform better at the high dose level with which the resolution would not be limited by image noises as much. This was not the case shown by our results for the large MCs. Furthermore, the edge of the small pixel size for the SSDM became insignificant while the tissue structure was present. This may be because of anatomic noise. Therefore, the smaller pixel size of the SSDM system may play an important but not significant role in clinical condition (tissue structure background and high dose of $1.74\ \text{mGy}$).

There are several limitations for this observer performance study. First is the image processing algorithms used to optimize image quality (or "for presentation"). Most of the processing tools are developed and provided by manufacturers, and are not well known to users. Therefore, it is not easy to apply similar processing algorithms to all images acquired with two different imaging units. Second, two separated display devices were used for reviewing the FFDM and the SSDM images. Both high-resolution CRT monitors employed P45 phosphor to display an image and had similar pixel size ($0.144\ \text{mm}$ versus $0.148\ \text{mm}$) and matrix size for display (2048×2560). Therefore, one can assume that the performances between these two CRT monitors are equivalent. For mammographic display, a full mammogram is always displayed for viewing. In this study, the matrix size of the FFDM images was 1914×2294 and 4096×5625 for the SSDM images. Therefore, the ratio of display size to the SSDM images was about 1:2 (50% size for viewing), which was not full-resolution view compared to the image display for the FFDM images. Further study is needed to investigate the effect of full- and half-resolution for image display. Third, there are several commercial FFDM systems which employ an anti-scatter grid method to reject scatter as well. With improved MTF and DQE for digital detectors and pro-

cessing algorithms for image display, these FFDM units provide high quality mammographic images and, thus, one may have different findings while comparing to the SSDM system.

V. CONCLUSION

The results from this study indicate that the CCD-based slot scanning DM system performed similarly to the FP-based area beam DM system with the uniform and the tissue structure backgrounds at both dose levels.

ACKNOWLEDGMENTS

This work was supported in part by research grant (No. EB-00117) from the National Institute of Biomedical Imaging and Bioengineering, a research grant (No. CA104759) from the National Cancer Institute, and a research grant (No. DADM17-00-1-0316) from the U.S. Department of Army Breast Cancer Research Program.

- ^{a)} Author to whom correspondence should be addressed. Telephone: (713) 745-2837. Electronic mail: chaolai@di.mdacc.tmc.edu
- ¹ J. V. Lacey, Jr., S. S. Devesa, and L. A. Brinton, "Recent trends in breast cancer incidence and mortality," *Environ. Mol. Mutagen.* **39**, 82–88 (2002).
 - ² American Cancer Society, Cancer Facts and Figures 2007, Atlanta, GA, American Cancer Society (2006).
 - ³ M. M. Tesic, M. Fisher Piccaro, and B. Munier, "Full field digital mammography scanner," *Eur. J. Radiol.* **31**, 2–17 (1999).
 - ⁴ S. Vedantham *et al.*, "Mammographic imaging with a small format CCD-based digital cassette: Physical characteristics of a clinical system," *Med. Phys.* **27**, 1832–1840 (2000).
 - ⁵ X. J. Rong *et al.*, "Microcalcification detectability for four mammographic detectors: Flat-panel, CCD, CR, and screen/film," *Med. Phys.* **29**, 2052–2061 (2002).
 - ⁶ M. J. Yaffe and J. A. Rowlands, "X-ray detectors for digital radiography," *Phys. Med. Biol.* **42**, 1–39 (1997).
 - ⁷ J. G. Mainprize, N. L. Ford, S. Yin, T. Tumer, and M. J. Yaffe, "Image quality of a prototype direct conversion detector for digital mammography," *Proc. SPIE* **3659**, 398–406 (1999).
 - ⁸ S. Vedantham *et al.*, "Full breast digital mammography with an amorphous silicon-based flat panel detector: Physical characteristics of a clinical prototype," *Med. Phys.* **27**, 558–567 (2000).
 - ⁹ S. Obenauer *et al.*, "Screen film vs full-field digital mammography: Image quality, detectability and characterization of lesions," *Eur. Radiol.* **12**, 1697–1702 (2002).
 - ¹⁰ S. Suryanarayanan *et al.*, "Flat-panel digital mammography system: Contrast-detail comparison between screen-film radiographs and hardcopy images," *Radiology* **225**, 801–807 (2002).
 - ¹¹ A. R. Cowen, J. H. Launder, M. Jadav, and D. S. Brett, "Visibility of microcalcifications in computed and screen-film mammography," *Phys. Med. Biol.* **42**, 1533–1548 (1997).
 - ¹² X. Liu, C. C. Shaw, X. J. Rong, and M. R. Lemacks, "Comparison of a-Si:H/CsI flat-panel digital imaging systems with CR and CCD based systems - image quality measurements," *Proc. SPIE* **4320**, 389–398 (2001).
 - ¹³ S. Arakawa, H. Yasuda, K. Kohda, and T. Suzuki, "Improvement of image quality in CR mammography by detection of emissions from dual sides of an imaging plate," *Proc. SPIE* **3977**, 590–600 (2000).
 - ¹⁴ J. A. Seibert, J. M. Boone, V. N. Cooper III, and K. K. Lindfors, "Cassette-based digital mammography," *Technol. Cancer Res. Treat.* **3**, 413–427 (2004).
 - ¹⁵ D. R. Dance and G. J. Day, "The computation of scatter in mammography by Monte Carlo methods," *Phys. Med. Biol.* **29**, 237–247 (1984).
 - ¹⁶ R. Fahrig, J. G. Mainprize, N. Robert, A. Rogers, and M. J. Yaffe, "Performance of glass fiber antiscatter devices at mammographic energies," *Med. Phys.* **21**, 1277–1282 (1994).
 - ¹⁷ J. M. Boone, K. K. Lindfors, V. N. Cooper III, and J. A. Seibert, "Scatter/primary in mammography: Comprehensive results," *Med. Phys.* **27**, 2408–2416 (2000).
 - ¹⁸ J. M. Boone, J. A. Seibert, C. M. Tang, and S. M. Lane, "Grid and slot scan scatter reduction in mammography: Comparison by using Monte Carlo techniques," *Radiology* **222**, 519–527 (2002).
 - ¹⁹ D. R. Dance, J. Persliden, and G. A. Carlsson, "Calculation of dose and contrast for two mammographic grids," *Phys. Med. Biol.* **37**, 235–248 (1992).
 - ²⁰ P. S. Rezendes, A. de Almeida, and G. T. Barnes, "Mammography grid performance," *Radiology* **210**, 227–232 (1999).
 - ²¹ Z. Jing, W. Huda, and J. K. Walker, "Scattered radiation in scanning slot mammography," *Med. Phys.* **25**, 1111–1117 (1998).
 - ²² S. Z. Shen, A. K. Bloomquist, G. E. Mawdsley, M. J. Yaffe, and I. Elbakri, "Effect of scatter and an antiscatter grid on the performance of a slot-scanning digital mammography system," *Med. Phys.* **33**, 1108–1115 (2006).
 - ²³ S. Suryanarayanan, A. Karellas, and S. Vedantham, "Physical characteristics of a full-field digital mammography system," *Nucl. Instrum. Methods Phys. Res. A* **533**, 560–570 (2004).
 - ²⁴ G. M. Besson *et al.*, "Design and evaluation of a slot-scanning full-field digital mammography system," *Proc. SPIE* **4682**, 457–468 (2002).
 - ²⁵ Z. Jing, W. Huda, J. K. Walker, and W. Y. Choi, "Detective quantum efficiency of a CsI:Tl scintillator based scanning slot x-ray detector for digital mammography," *Proc. SPIE* **3336**, 583–591 (1998).
 - ²⁶ D. R. White, "The formulation of tissue substitute materials using basic interaction data," *Phys. Med. Biol.* **22**, 889–899 (1977).
 - ²⁷ C. B. Caldwell and M. J. Yaffe, "Development of an anthropomorphic breast phantom," *Med. Phys.* **17**, 273–280 (1990).
 - ²⁸ C.-J. Lai *et al.*, "Receiver operating characteristic analysis for the detection of simulated microcalcifications on mammograms using hardcopy images," *Phys. Med. Biol.* **51**, 3901–3919 (2006).
 - ²⁹ X. Wu, E. L. Gingold, G. T. Barnes, and D. M. Tucker, "Normalized average glandular dose in molybdenum target-rhodium filter and rhodium target-rhodium filter mammography," *Radiology* **193**, 83–89 (1994).
 - ³⁰ National Council on Radiation Protection and Measurements, "Mammography: A user's guide: Recommendations of the National Council on Radiation Protection and Measurements Chapter 3: Dose evaluation," NCRP Report No. 85, Bethesda, MD, 1986.
 - ³¹ C.-J. Lai *et al.*, "A contrast-detail phantom study to compare a slot scanning digital mammography system with an a-Si/CsI flat-panel based system," *Med. Phys.* **31**, 1796 (2004).
 - ³² D. D. Dorfman, K. S. Berbaum, and C. E. Metz, "Receiver operating characteristic rating analysis: Generalization to the population of readers and patients with the jackknife method," *Invest. Radiol.* **9**, 723–731 (1992).
 - ³³ D. D. Dorfman *et al.*, "Monte Carlo validation of a multireader method for receiver operating characteristic discrete rating design: Factorial experimental design," *Acad. Radiol.* **5**, 591–602 (1998).
 - ³⁴ S. L. Hillis and K. S. Berbaum, "Power estimation for the Dorfman-Berbaum-Metz method," *Acad. Radiol.* **11**, 1260–1273 (2004).
 - ³⁵ S. L. Hillis and K. S. Berbaum, "Monte Carlo validation of the Dorfman-Berbaum-Metz method using normalized pseudovalues and less data-based model simplification," *Acad. Radiol.* **12**, 1534–1541 (2005).
 - ³⁶ S. L. Hillis, N. A. Obuchowski, K. M. Schartz, and K. S. Berbaum, "A comparison of the Dorfman-Berbaum-Metz and Obuchowski-Rockette methods for receiver operating characteristic (ROC) data," *Stat. Med.* **24**, 1579–1607 (2005).
 - ³⁷ X. Liu, C. C. Shaw, M. C. Altunbas, and T. Wang, "An alternate line erasure and readout (ALER) method for implementing slot-scan imaging technique with a flat-panel detector—initial experiences," *IEEE Trans. Med. Imaging* **25**, 496–502 (2006).