Effects of exposure equalization on image signal-to-noise ratios in digital mammography: A simulation study with an anthropomorphic breast phantom

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Purpose: The scan equalization digital mammography (SEDM) technique combines slot scanning and exposure equalization to improve low-contrast performance of digital mammography in dense tissue areas. In this study, full-field digital mammography (FFDM) images of an anthropomorphic breast phantom acquired with an anti-scatter grid at various exposure levels were superimposed to simulate SEDM images and investigate the improvement of low-contrast performance as quantified by primary signal-to-noise ratios (PSNRs).

Methods: We imaged an anthropomorphic breast phantom (Gammex 169 “Rachel,” Gammex RMI, Middleton, WI) at various exposure levels using a FFDM system (Senographe 2000D, GE Medical Systems, Milwaukee, WI). The exposure equalization factors were computed based on a standard FFDM image acquired in the automatic exposure control (AEC) mode. The equalized image was simulated and constructed by superimposing a selected set of FFDM images acquired at 2, 1, 1/2, 1/4, 1/8, 1/16, and 1/32 times of exposure levels to the standard AEC timed technique (125 mAs) using the equalization factors computed for each region. Finally, the equalized image was renormalized regionally with the exposure equalization factors to result in an appearance similar to that with standard digital mammography. Two sets of FFDM images were acquired to allow for two identically, but independently, formed equalized images to be subtracted from each other to estimate the noise levels. Similarly, two identically but independently acquired standard FFDM images were subtracted to estimate the noise levels. Corrections were applied to remove the excess system noise accumulated during image superimposition in forming the equalized image. PSNRs over the compressed area of breast phantom were computed and used to quantitatively study the effects of exposure equalization on low-contrast performance in digital mammography.

Results: We found that the highest achievable PSNR improvement factor was 1.89 for the anthropomorphic breast phantom used in this study. The overall PSNRs were measured to be 79.6 for the FFDM imaging and 107.6 for the simulated SEDM imaging on average in the compressed area of breast phantom, resulting in an average improvement of PSNR by ~35% with exposure equalization. We also found that the PSNRs appeared to be largely uniform with exposure equalization, and the standard deviations of PSNRs were estimated to be 10.3 and 7.9 for the FFDM imaging and the simulated SEDM imaging, respectively. The average glandular dose for SEDM was estimated to be 212.5 mrad, ~34% lower than that of standard AEC-timed FFDM (323.8 mrad) as a result of exposure equalization for the entire breast phantom.

Conclusions: Exposure equalization was found to substantially improve image PSNRs in dense tissue regions and result in more uniform image PSNRs. This improvement may lead to better low-contrast performance in detecting and visualizing soft tissue masses and micro-calcifications in dense tissue areas for breast imaging tasks. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3659709]

Key words: digital mammography, exposure equalization, contrast-to-noise ratio, low-contrast performance, average glandular dose

I. INTRODUCTION

Breast cancer is one of the leading causes of cancer death in women. Approximately one in eight women in the United States will develop invasive breast cancer in her lifetime.¹ Early detection of breast cancer is critical for the patient's survival. For the past several decades, mammography has been known to be the most sensitive and efficient method for detecting early-stage breast cancer. The higher sensitivity and effectiveness of mammography were due in part to the use of the high-contrast screen-film (SF) system, which enables low-contrast lesions to be differentiated and visualized. However, the high-contrast SF combination imposes a narrow dynamic range, which limits the useful exposure range during mammographic procedures and results in a large amount of breast tissue being imaged with a low-contrast appearance. This problem becomes even more severe when a larger and denser breast is examined. Women with the highest levels of breast density have a risk of breast cancer four to six times that of women with the lowest level of breast density.²⁻⁴

To overcome the shortcoming of the narrow latitude associated with the high-contrast SF system, a variety of exposure equalization techniques have been proposed and investigated to improve image quality in mammography. One such method, which is based on an area beam equalization approach, uses a solid or elastic beam filter/attenuator to compensate for overexposure either around the breast periphery only or throughout the entire breast.⁵⁻¹³ Adding such a compensator to the breast periphery is the simplest way to improve visualization at the breast periphery of the SF system. However, this technique can be difficult to implement, especially for oblique views. Another problem with the compensator approach is that the added attenuating material may degrade image contrast in the area near the breast periphery by increasing scattered radiation to the breast tissue. The full-area beam equalization method compensates for overexposures throughout the entire breast rather than at the breast periphery only.¹⁴ Although full-area beam equalization method provides the benefits of short x-ray exposure time and a low x-ray tube heat loading that is comparable to that of conventional mammography, it has disadvantages. In addition to increased scattered radiation from the beam compensator, this technique can be difficult to implement because of misalignment and because it takes a relatively long time to construct and position a patient-specific filtering mask.

Another, more sophisticated method of exposure equalization involves scanning the breast with an x-ray pencil beam or fan beam, whose intensity or width is modulated regionally according to feedback from the regional variations in x-ray transmission as measured by the image detector.¹⁵⁻²² This method can equalize the detector exposure throughout the entire breast without adding any beam-filtering material and can achieve a largely uniform signal-to-noise ratio (SNR). In addition, the use of a slot-shaped aft-collimator reduces scattered radiation substantially without attenuating the primary x-rays, hence improving image

quality without sacrificing dose efficiency. However, this method requires a longer exposure time and a higher x-ray tube heat loading than those of conventional mammography. Previous advances in scanning equalization mammography have been implemented and investigated to overcome the narrow latitude of the SF system. Some exposure equalization techniques have been initially demonstrated in clinical practice and evaluated by means of various phantoms, with encouraging results.^{16,20,21}

Over the past two decades, the developments in advanced electronics and computer technology have led digital detection technology to gain rapid acceptance in radiology departments and to gradually replace SF systems.²³⁻³⁷ Digital detectors can overcome certain fundamental limitations associated with SF systems by providing a wide dynamic range and enabling contrast enhancement and the post-processing of images. In digital format, image contrast can be freely manipulated by varying the window/level. However, image contrast alone does not determine the low-contrast performance because image noise also plays an important role in perceiving low-contrast objects in digital images. In other words, a higher noise level may prevent abnormal soft tissue masses or micro-calcifications from being detected or visualized in dense areas of the breast. The exposure equalization can improve the image SNR and low-contrast performance by compensating for x-ray intensities in dense areas of the breast in digital mammography. In addition, the exposure equalization can reduce average glandular dose by minimizing the radiation exposure to the breast periphery.²²

Recently, we developed a scan equalization digital radiography (SEDR) system that combines slot-scan geometry, regional beam width modulation, and digital detection technology to reduce scattered radiation and to improve SNRs in heavily attenuating regions of the chest.^{38,39} An electronic aft-collimation technique, referred to as the alternate line erasure and readout (ALER) technique, was implemented with a flat-panel (FP) detector to achieve scatter reduction without using a large and heavy aft-collimator.^{40,41} The same technique can be applied to digital mammography to improve SNRs of soft tissue masses and the visibility of micro-calcifications in dense areas of the breast. Here, we performed a simulation study using an anthropomorphic breast phantom to explore and investigate the benefits of scan equalization in digital mammography. Although the proposed scan equalization digital mammography (SEDM) system is still under construction, we have used images acquired with a full-field digital mammography (FFDM) system to simulate exposure equalization and to investigate its effects on low-contrast performance and average glandular dose in digital mammography.

II. MATERIALS AND METHODS

II.A. Imaging system

We used a cesium iodide/amorphous silicon (CsI:TI/a-Si) indirect-conversion flat-panel (FP) based full-field digital mammography (FFDM) system (Senographe 2000D,

General Electric Medical Systems, Milwaukee, WI) to acquire images in this study. The imaging system was manufactured as a stand-alone unit and integrated with a linear anti-scatter grid (5:1 grid ratio; 31 lines/cm). The active area of the detector is divided into a 1914×2294 array of image elements, each with a pixel size of $100 \mu\text{m}$. The imaging system provides both “raw” and “processed” image data with 14-bit depth. The “raw” image data are pre-processed for dark current, gain, and dead pixel correction, and they are linear to the detector exposure. They are usually used for physical image quality metrics, SNR, and contrast-to-noise ratio (CNR) measurement. The “processed” image data are logarithmic transformed from the “raw” image data for archiving, soft-copy displaying, and hard-copy printing. The upper surface of the image receptor (which acts as the breast support) is a removable Bucky with the anti-scatter grid housed in a protective enclosure. The removable Bucky can be replaced with an optional breast support without a grid when the grid is not required, for instance, in the case of a needle biopsy performed in the magnification mode. The details of the FFDM imaging system along with physical imaging characteristics and perception performance have been previously investigated and reported.^{29,35,37,42}

Before acquiring images, we performed a complete system calibration (correction for dark current bias, gain, and defective pixels/lines) according to the manufacturer’s guidelines. The source-to-image distance (SID) was maintained at 66 cm. The half-value layer (HVL) was measured to be 0.37 mm aluminum (Al) at 28 kVp with the molybdenum/molybdenum (Mo/Mo) target/filter combination used in this study. Raw image data were archived onto CD-R disks and then transferred to a workstation for further image processing.

II.B. Anthropomorphic breast phantom

An anthropomorphic breast phantom (Gammex 169 “Rachel,” Gammex RMI, Middleton, WI) was used to mimic a “regular size breast” and to evaluate the low-contrast performance of the SEDM technique. Unlike commonly used tissue-equivalent test phantoms (which are usually uniform in density with different thicknesses and adipose/glandular compositions), the anthropomorphic breast phantom is a unique testing instrument that produces an image with contrast range and details comparable to those of clinical mammograms. The active portions of the phantom consist of a base and an imaging target. The base is constructed using breast-equivalent material (BR-12) to simulate typical breast density. Unlike a real compressed breast to be imaged (with fixed thickness and variable tissue densities), the breast phantom uses a single composition material to mimic the attenuation factor via thickness variation. Therefore, the scatter distribution in the breast phantom may be different from that of a real breast. The anthropomorphic breast phantom was positioned with the help of a custom-built Al frame fixture to ensure the precision of repositioning throughout the study. We acquired breast phantom images at various x-ray exposure levels and a fixed tube potential of 28 kVp

with a Mo/Mo target/filter combination. The system setup is shown in Fig. 1.

II.C. Characterization of system linearity and noise level

System linearity is essential for the measurement of various image quality metrics, such as the modulation transfer function (MTF), noise power spectra (NPS), detective quantum efficiency (DQE), SNR, and contrast ratio (CR). To assess system linearity and noise levels, we acquired a series of flat-field images using a 50-mm-thick polymethyl methacrylate (PMMA) slab (with a 50% glandular/50% adipose composition) to simulate x-ray attenuation by a typical compressed breast. The PMMA slab was placed on the breast compression paddle at 20 cm from the focal spot to minimize scattered radiation and to produce nearly uniform exposure at the detector. The breast compression paddle release feature (decompression after each exposure) was disabled to maintain the position of the PMMA slab. Exposures were made at various mAs settings (ranging 4–500 mAs). Measurements were performed with and without the anti-scatter grid in place over a $50 \times 50 \text{ mm}^2$ area (20 mm from the chest wall). The area was divided into 10×10 regions-of-interest (ROIs), each containing 50×50 pixels. The mean pixel value and standard deviation were measured for each ROI and then averaged over all ROIs. The averaged mean pixel values were then plotted as a function of detector exposure to check the system linearity. For a perfect FP detector system, the mean pixel value, PV has a linear relationship with the detector exposure X (in mAs), as follows:

$$PV = a_0 + k \cdot X, \quad (1)$$

where a_0 is the dark current bias, k is the exposure-to-pixel-value conversion factor, a_0 can be minimized through the detector dark current correction (ideally, a_0 should equal zero).

We measured the noise level in the flat-field image by computing the standard deviation of the pixel value over the ROIs. The image noise is assumed to consist of three major components: system noise, quantum (x-ray photon) noise, and structural noise. The variance of the total image noise

Fig. 1. The anthropomorphic breast phantom “Rachel” was imaged using a GE Senographe 2000D FFDM system.

can be expressed as the sum of the variances of these three components as follows:⁴³

$$\sigma_t^2 = \sigma_e^2 + \sigma_q^2 + \sigma_s^2 = a + b \cdot X + c \cdot X^2, \quad (2)$$

where σ_e^2 , σ_q^2 , and σ_s^2 are the variances of the system noise, quantum noise, and structural noise, respectively, a is a constant that represents the variance of the system noise, and b and c are scaling factors for the quantum and structural noises, respectively. A second-order polynomial function can be used to fit the plot of measured noise variance versus detector exposure using the least-squares method to determine the coefficients a , b , and c and the variance of each noise component. For clinical applications, the quantum noise is expected to be the dominant component. The exposure X can be replaced with mean pixel value if the detector system is well calibrated (of course, with different constant a and scaling factors b and c). The characterization of the noise level can provide useful insight into the properties of each noise component in the simulation of SEDM imaging.

II.D. Determination of slot width in SEDM

One feature of SEDM is the use of slot-scan geometry to reduce scattered radiation while keeping the primary x-rays unchanged. To determine the effectiveness of scatter reduction with the SEDM technique, the scatter-to-primary ratio (SPR) has to be measured with different slot widths and compared with that of the anti-scatter grid method in conventional mammography. We used an edge device technique³⁸ to measure the SPR along a line parallel to one of the detector sides for both slot-scanning and FFDM. With this method, a rectangular tungsten plate (0.4 mm thick and 30 mm wide, long enough to cover the long side of the FP detector) with a straight and sharp edge was placed directly on top of the breast phantom. The tungsten plate was carefully positioned so that the sharp edge was aligned parallel to the chest wall and image lines at 30 mm from the chest wall. The distance between the tungsten plate and the detector surface was maintained at 10 cm. The penumbra effect caused by focal spot blurring, estimated to be 0.05 mm on the image plane for the large focal spot (0.3 mm nominal), was negligible in the scatter measurement. The 0.4-mm-thick tungsten plate blocked nearly all primary x-rays for mammographic spectra, leaving only the scattered radiation under the blocked area.

Two images were acquired to estimate the scattered radiation along the edge of the tungsten plate. The first image was acquired with the front portion (against the chest wall) of the breast phantom blocked by the tungsten plate, and the second image was acquired with the edge re-oriented 180° from its original position with the back portion (away from the chest wall) of the breast phantom blocked by the tungsten plate. An additional large piece of tungsten plate was used to block a large portion of the phantom for the second image. The scatter distribution $I_S(x)$ was estimated by summing the scatter components from the two measurements along the shadow of the edge. A third breast phantom image was acquired without the

presence of the tungsten plate to measure the total (primary plus scatter) image signal $I_{P+S}(x)$ at the position where the scatter signal had been measured. The primary signal, $I_P(x)$, was computed as the difference between the total and scatter signals, $I_{P+S}(x) - I_S(x)$. After $I_S(x)$ and $I_{P+S}(x)$ were measured, the position-dependent SPR, $SPR(x)$, was computed as follows:

$$SPR(x) = \frac{I_S(x)}{I_P(x)} = \frac{I_S(x)}{I_{P+S}(x) - I_S(x)}. \quad (3)$$

The position-dependent SPRs were measured with the edge device technique for the FFDM images acquired with and without the anti-scatter grid, and for the slot-scanning images acquired with slot widths of 5, 10, 15, and 20 mm (no grid). The slot-scan images were acquired by exposing the breast phantom to a narrow fan-beam x-ray at the location where position-dependent SPRs were measured. A pair of rectangular 0.4-mm-thick tungsten plates was placed on top of the breast phantom to form a narrow slot. The gap opening of the tungsten plates was aligned and collimated carefully with the detector edge to form slot widths of 5, 10, 15, and 20 mm, respectively, on the image plane. The position-dependent SPRs were measured along the central line of the slot. The measured position-dependent SPRs were used to help select a proper slot width for simulation of SEDM.

II.E. Assessment of primary and scattered radiation

The pinhole aperture array method was used to estimate the spatial distribution of the scattered radiation and the primary component at any of sampling point in the breast phantom image and to assess the low-contrast performance and average glandular dose in SEDM.⁴¹ With this method, a 1-mm-thick lead sheet (laminated between two 1-mm-thick acrylic sheets) containing an array of pinhole apertures, each 1 mm in diameter with a pitch of 10 mm, was placed directly on top of the anthropomorphic breast phantom and imaged at 28 kVp and 125 mAs with a Mo/Mo target/filter combination. The scatter components in the pinhole aperture areas can be estimated by acquiring two breast phantom images with and without the pinhole aperture mask in place as follows:

$$\begin{aligned} I_S(i,j) &= I_{P+S}(i,j) - I_P(i,j) \\ &\approx I_{P+S}(i,j) - I_{Aperture}(i,j), \end{aligned} \quad (4)$$

where (i,j) is the coordinate of the pinhole aperture location, $I_{P+S}(i,j)$ is the primary plus scatter signal measured in the image acquired without the pinhole aperture mask, and $I_{Aperture}(i,j)$ is the signal measured within the pinhole aperture, is an approximation for $I_P(i,j)$ because $I_{Aperture}(i,j)$ still contains a small amount of scatter component resulting from volumes irradiated by the narrow pencil beams and cross-talk from neighboring apertures. The approximation yields an overestimates of the primary signal by ~1.2% (0.8% from in-aperture scatter and 0.4% from cross-talk), according to a method described in a previous study.⁴¹

To improve the accuracy of the scatter estimation with the pinhole aperture array method, we acquired four images separately and imaged each with the lead aperture mask plate shifted by 5 mm from right to left or from front to back, resulting in a pinhole sampling density increased by a factor of four compared to that of one image alone. The image signals of the central 3×3 pixels were averaged in each aperture location to estimate the primary signal (from the approximation $I_{\text{Aperture}}(i, j)$) and the primary plus scatter signal $I_{P+S}(i, j)$ for a total of 18×38 pinhole locations (9×19 in each masked breast phantom image), which covered the entire breast phantom area. The scatter estimates $I_S(i, j)$ were computed from Eq. 4, and the pixel-wise scatter signals in the entire breast phantom image can be estimated by fitting a smooth surface to the scatter estimates $I_S(i, j)$ at the 18×38 pinhole locations. We used the following quadratic surface function of the spatial location (x, y) to estimate the pixel-wise spatial distribution of scattered radiation in the entire breast phantom image

$$I_S(x, y) = a_0 + a_1 \cdot x + a_2 \cdot x^2 + a_3 \cdot x \cdot y + a_4 \cdot y + a_5 \cdot y^2. \quad (5)$$

We used a least-squares method to determine the coefficients of the scatter function. The pixel-wise scatter fraction SF and the primary signal can be computed as follows:

$$SF(x, y) = \frac{I_S(x, y)}{I_{P+S}(x, y)}, \quad (6)$$

$$I_P(x, y) = I_{P+S}(x, y) \cdot (1 - SF(x, y)). \quad (7)$$

II.F. Simulation of exposure equalization in digital mammography

The objective of exposure equalization is to regionally modulate the exposure level and to achieve coarsely (averaged over the entire breast phantom area) uniform detector exposure and image SNR, while keeping the average entrance skin exposure (ESE) at a level comparable to that for standard digital mammography. A SEDM system may be implemented with a similar technique we used previously with the SEDR imaging^{37–40} to improve the visibility of micro-calcifications in denser tissue areas and the low-contrast performance of soft tissue masses in digital mammography. To achieve exposure equalization, a narrow fan-beam x-ray (oriented parallel to the anode–cathode axis) can be used to scan the breast phantom in the direction perpendicular to the anode–cathode axis during the x-ray exposure. The x-ray exposures to the image detector can be regionally controlled in such a way that lower exposure intensity is expected for less attenuating areas (fatty tissues or breast periphery) and higher exposure intensity is expected for highly attenuating areas (denser tissue masses) by modulating either the tube current or the regional beam width. A multiple-segment beam width modulator array can be built to modulate the x-ray fan-beam width regionally during the x-ray exposure. A schematic drawing of the proposed SEDM is illustrated in Fig. 2.

On the basis of the SPR measurement performed in Sec. II D, we used FFDM images acquired with the anti-scatter grid at various exposure levels to simulate the SEDM imaging technique with a slot width of 15 mm. To start with, we

FIG. 2. Schematic drawing of the proposed scanning equalization digital mammography (SEDM) system.

used a standard FFDM image of the breast phantom acquired in AEC mode (125 mAs and 28 kVp) as a pre-scan image to determine the exposure equalization factor. First, the FFDM image was cropped to a size of 980×2080 pixels, which covers the entire breast area. Image signals were then rescaled by multiplying a factor of $1/T_p$ (where T_p is the primary transmission of the anti-scatter grid) in order to keep the primary signal levels the same as those acquired with the slot-scan imaging technique. Second, image lines of the breast phantom were divided evenly into 24 segments in the anode–cathode axis direction, each with 50 image lines (or 5 mm wide), corresponding to a small segment covered by a beam modulation blocker. The segments overlapped by 10 image lines on each side to blur the edges and to produce a smooth transition between two neighboring segments. Image data were then averaged along the anode–cathode axis direction within each segment to yield 24 mean signal profiles in the fan-beam scanning direction. These signal profiles were convolved with a low-pass filtering kernel to smooth out their shape and to reduce the noise fluctuations. The exposure equalization factor used to regulate the entrance skin exposure for each individual segment can be computed as follows:

$$E_{eq}(i,j) = \frac{\overline{I_{pre}}}{I_{pre}(i,j)}, \quad (8)$$

where the subscripts *eq* and *pre* refer to “equalization” and “pre-scan,” respectively, *i* and *j* refer to segment number and row number, respectively, and $\overline{I_{pre}}$ is the average image signal computed over the compressed area of the breast phantom (pixels with intensity >4 times the minimum intensity were excluded from the calculation of the average image signal). The exposure equalization factors should be smaller than unit for less dense areas, such as fatty tissue or the periphery of the breast, indicating an exposure reduction. The exposure equalization factors should be greater than unit for denser tissue areas, indicating an increase in exposure to compensate for the higher x-ray attenuation. The exposure equalization factors computed for the simulation of SEDM are shown in Fig. 3(a).

The exposure equalization factors computed from Eq. (8) can be used to modulate the beam width modulation array regionally and to equalize the detector input exposure for a real SEDM system using a method similar to that we used previously in a SEDR system.³⁸ The product of the exposure equalization factors and the exposure for pre-scan imaging (125 mAs multiplied by $1/T_p$) can be used to determine the regional exposure levels needed at various areas in the SEDM image as follows:

$$mAs_{SEDM}(i,j) = E_{eq}(i,j) \cdot \frac{125}{T_p}. \quad (9)$$

A complete set of FFDM images acquired at various exposure levels (e.g., with 1 mAs increments) that covers the whole exposure range can be used to generate an SEDM image; however, the selection of the exposure levels (mAs)

FIG. 3. (a) Exposure equalization factors computed from a pre-scan FFDM image of the anthropomorphic breast phantom; (b) simulated SEDM image; (c) restored SEDM image.

is limited with commercial mammography systems. An alternative approach is to acquire a limited number of FFDM images at preselected exposure levels. The exposure from a single FFDM image or the sum of several FFDM images can be used to make up the exposure level determined by Eq. (9). Therefore, the number of FFDM images required to superimpose an SEDM image could be reduced markedly. The superimposed SEDM image can be computed as follows:

$$I_{SEDM}(i,j) = \sum_{k=1}^7 a_k(i,j) \cdot I_k(i,j), \quad (a_k = 0, 1), \quad (10)$$

where the coefficient a_k is determined by the exposure equalization factors and I_k is the signal intensity for FFDM images acquired at 1/32, 1/16, 1/8, 1/4, 1/2, 1, and 2 times the exposure level of the standard AEC timed image (125 mAs in this study). This approach is similar to that used in the binary system (to use seven binary bits to represent decimals between 0 and 127). In this study, we used the FFDM images acquired from the following pre-selected exposure (mAs) levels to superimpose an SEDM image: 4, 8, 16, 32, 63, 125, and 250 mAs. Images from the last three exposures were scaled up to have a signal intensity, the same as that of images acquired at 64, 128, and 256 mAs. Errors of signal intensity in the simulated SEDM image are expected to be ± 2 mAs (or $\pm 2\%$). The superimposed SEDM image of the anthropomorphic breast phantom demonstrates coarsely uniform exposure levels over the compressed tissue area, as shown in Fig. 3(b). The SEDM image can be restored, by dividing by the exposure equalization factors, to have the appearance of a standard mammogram, as shown in Fig. 3(c), with image inversion and logarithmic transformation required. The restoration process (without inversion and logarithmic transformation) does not alter the image; hence, we used the restored SEDM image to assess the low-contrast performance. The major steps in the simulation of exposure equalization are shown in Fig. 4.

Fig. 4. Flow chart to illustrate the process of simulating exposure equalization.

II.G. Assessment of low-contrast performance

The contrast signal in a digital image is defined as the signal difference between the low-contrast object and the area surrounding it and can be approximated as follows:

$$I - I' = (I_P + I_S) - (I'_P + I'_S) = I_P - I'_P = I_P \cdot (1 - e^{-\Delta\mu \cdot \Delta t}) \approx I_P \cdot (\Delta\mu \cdot \Delta t), \quad (11)$$

where $\Delta\mu$ is the difference in the linear attenuation coefficient between the low-contrast object and the background and Δt is the thickness of the low-contrast object. As $\Delta\mu \cdot \Delta t$ does not change with the scatter rejection technique (anti-scatter grid or slot-scan), the CNR is approximately proportional to the primary signal-to-noise ratio

$$CNR = \frac{I - I'}{\sigma} \approx \frac{I_P}{\sigma} \cdot \Delta\mu \Delta t \propto \frac{I_P}{\sigma}. \quad (12)$$

Thus, we can use the primary signal-to-noise ratios (PSNRs) to compare the low-contrast performance of various imaging systems or techniques as long as the object contrast, $\Delta\mu \Delta t$, does not change (generally the case if the tube potential and target/filter combination remain the same). The advantage of using the PSNRs for low-contrast performance comparison is that imaging of a low-contrast object is not required.

Two sets of FFDM images with identical technical settings were acquired, to superimpose the SEDM images, and then subtracted from each other to compute the noise level. The process removed the anatomical structures and other structural noises, leaving a noise-only image. This process was also applied to the FFDM images acquired at 125 mAs. Proper normalization was performed before the subtraction

of two images to minimize the residual structural noise that may have arisen from exposure fluctuations. The subtracted images were divided into sub-regions, each with 40×40 pixels, over which the standard deviations of image data were computed and divided by $\sqrt{2}$ to estimate the noise levels within each sub-region. The primary signal was averaged within each sub-region as well.

The standard deviation computed from the subtracted SEDM images cannot be directly used to estimate the true image noise. The image variance in the superimposed SEDM, using the method described by Eq. (10), can be expressed as

$$\sigma_{SEDM}^2 = a \cdot \sum_{k=1}^7 a_k + b \cdot \sum_{k=1}^7 a_k \cdot X_k + c \cdot \sum_{k=1}^7 a_k \cdot X_k^2, \quad (a_k = 0, 1). \quad (13)$$

Because the structural noise remained the same in all images as long as the signal level was the same, the structural noise component (the third term) canceled out during the subtraction of two SEDM images. The quantum noise component (the second term) remained in a linear relationship with the total detector exposure, as expected. However, the level of the system noise component (the first term) could increase with the number of FFDM images (ranging from 1 to 7) that were superimposed in each region for exposure equalization. This resulted in a substantial increase of total noise variance in certain regions where many FFDM images had been used to construct the simulated SEDM images. Even though the FP imaging system is optimized with quantum noise as the dominant noise component, the increased system noise could become substantially high. Thus, this increase in system noise must be corrected for before the variance of the subtracted SEDM image can be used to assess the PSNR. The average number of FFDM images, n , used to construct the SEDM images for each sub-region was computed and used to correct the variance of the simulated SEDM image as follows:

$$\sigma_{Corrected}^2 = \sigma_{Subtraction}^2 / 2 - a \cdot (n - 1), \quad (14)$$

where a is the system noise variance, as in Eq. (2). For regions where only a single FFDM image was used, the excess system noise [the second term in Eq. (14)] becomes zero and the total image noise equals the standard deviation of the subtracted image divided by $\sqrt{2}$. The largest number of FFDM images used to construct the SEDM image was found to be 6 in our simulation. After performing the noise correction, we used the primary signal from Eq. (7) to compute the PSNRs within each ROI (40×40 pixels) as follows:

$$PSNR(i, j) = \frac{I_P(i, j)}{\sigma_{Corrected}(i, j)}. \quad (15)$$

As the object contrast $\Delta\mu \Delta t$ remains the same in SEDM, the CNR improvement factors defined as CNRs of the SEDM image over the FFDM image with the grid can be estimated by computing the PSNR improvement factors (PSNRIFs) within each ROI (40×40 pixels) as follows:

$$PSNR_{IF}(i,j) = \frac{PSNR_{SEDM}(i,j)}{PSNR_{FFDM}(i,j)}. \quad (16)$$

II.H. Approximation of average glandular dose

Average glandular dose is often used as the index to indicate the radiation dose risk in breast imaging. By definition, the average glandular dose is computed as follows:⁴⁴

$$D_g = D_{gN} \cdot X_{ESE}, \quad (17)$$

where X_{ESE} is the ESE, and D_{gN} is the ESE-to-average-glandular-dose conversion factor. D_{gN} is determined by experimental and computer simulation methods and is dependent on the following factors: x-ray tube potential (kVp), beam quality (HVL), x-ray tube target/filter combination, breast thickness, and breast tissue composition. The average glandular dose for the entire anthropomorphic breast phantom can be calculated from the sum of all pixelated $D_g(i,j)$ values by following a previously reported method, as follows:¹⁴

$$D_g = \frac{\sum_{i=1}^I \sum_{j=1}^J D_g(i,j)}{I \cdot J} = \frac{\sum_{i=1}^I \sum_{j=1}^J D_{gN}(i,j) \cdot X_{ESE}(i,j)}{I \cdot J} = \frac{\sum_{i=1}^I \sum_{j=1}^J D_{gN}(i,j) \cdot mAs(i,j) \cdot C(i,j)}{I \cdot J}, \quad (18)$$

where $mAs(i,j)$ changes region-by-region in SEDM but remains a constant (125 mAs) in FFDM and $C(i,j)$ is the mAs-to-ESE conversion factor and may change region-by-region because of the x-ray heel effect. To simplify the process, we assumed ESE to be homogeneous in standard mammography, and $C(i,j)$ becomes a constant (11.2 mR/mAs measured at 5 cm above the breast support platform at 28 kVp and 0.37 mm Al HVL with a Mo/Mo target/filter combination). Thus, $X_{ESE}(i,j)$ is proportional to $mAs(i,j)$ only. To determine the regionally dependent $D_{gN}(i,j)$, we imaged a set of PMMA slabs of various thicknesses with a 50% glandular/50% adipose composition with a fixed exposure of 125 mAs at 28 kVp with a Mo/Mo target/filter combination. An experimental curve of primary intensity (measured with the pinhole aperture method) versus slab thickness was plotted and compared with the primary intensity of the FFDM breast phantom image to determine the spatial variation of the breast thickness. Finally, the spatially dependent $D_{gN}(i,j)$ can be computed using the derived breast thickness image and a curve of D_{gN} versus breast thickness interpolated from previously published data.⁴⁴ The relative glandular dose RD_g , defined as the glandular dose of SEDM over that of FFDM, was computed as follows:

$$RD_g(i,j) = \frac{D_g^{SEDM}(i,j)}{D_g^{FFDM}(i,j)}. \quad (19)$$

III. RESULTS

We characterized the system linearity by measuring the mean pixel values of flat-field images acquired at various ex-

posure levels and then plotting the values as a function of tube-current–exposure time product (mAs). The system linearity measured with and without the anti-scatter grid is shown in Fig. 5. Both curves were found to have a perfect linear relationship with exposure (with correlation coefficient $R^2 = 1$).

The variances of the total noise and individual noise components measured in the flat-field images acquired using the 50-mm-thick PMMA slab with the grid are plotted as a function of pixel value in Fig. 6(a). We found that the measured noise variance fits very well to the noise model (with correlation coefficient $R^2 = 1$) for the FP detector used in this study. The fractions of component variance, computed as each component variance over the total noise variance, are plotted and compared in Fig. 6(b). The fractional approach enables easy identification of the dominant noise source for a given signal range and imaging setup. We found that the system noise (additive) dominates at low signal levels (i.e., low detector exposure), whereas the structural component becomes increasingly important at higher signal levels (i.e., higher detector exposure). In addition, the quantum noise component remained dominant (equal or greater than 74% of total noise) for mean pixel values ranging from ~500 to 3543 (resulting from the highest allowable exposure of 500 mAs at 28 kVp with a Mo/Mo target/filter combination) and fall into the exposure range used clinically, as was expected for a well-designed FP detector system. For the image of the anthropomorphic breast phantom acquired in AEC mode (125 mAs at 28 kVp with a Mo/Mo target/filter combination), the minimum and mean pixel values were 787 and 1517, respectively, indicating that the detector exposure is optimized in AEC mode, where the quantum variance fractions were 80% and higher. For the SEDM image, the detector exposure was equalized to yield pixel values of ~2200, where the quantum variance fraction was dominant as well (nearly 80%).

The spatially dependent SPRs measured for slot-scan images acquired with various slot widths (5, 10, 15, and

FIG. 5. System linearity plotted as mean pixel value versus exposure (mAs) for images acquired with and without the anti-scatter grid. Exposures made at 28 kVp using a 50-mm-thick PMMA slab (50% adipose/50% glandular composition).

FIG. 6. (a) Variances of total noise and individual noise components plotted as a function of mean pixel value; (b) fraction of noise component variance plotted as a function of pixel value. Measurements made at 28 kVp with various exposure levels using a 50-mm-thick PMMA slab (50% adipose/50% glandular composition).

20 mm) and for FFDM images with and without the anti-scatter grid using the edge device technique are shown in Fig. 7 for comparison. The average SPRs (averaged between 35 and 195 mm) were measured to be 0.22 and 0.68 for FFDM with and without the grid, respectively. For slot-scan imaging, the average SPRs were 0.10, 0.17, 0.23, and 0.28 for slot widths of 5, 10, 15, and 20 mm, respectively. We found that both the anti-scatter grid and slot-scan methods reduced scattered radiation substantially. The anti-scatter grid reduced SPR by 68% on average, whereas the slot-scan method reduced SPR by 86%, 75%, 67%, and 59% for slot widths of 5, 10, 15, and 20 mm, respectively. The narrower the slot, the greater the reduction in SPR. We found that the 15-mm-wide slot resulted in approximately the same SPRs as achieved with the anti-scatter grid. Thus, we elected to use FFDM images acquired with the grid to simulate SEDM

FIG. 7. SPRs measured for slot-scan images acquired with various slot widths (5, 10, 15, and 20 mm) and for FFDM images with and without the anti-scatter grid.

with a maximum fan-beam width of 15 mm in Sec. II F. In addition, the slot width should be ≤ 15 mm to preserve scatter rejection ability comparable to that of the anti-scatter grid for building an SEDM imaging system.

In Figs. 8(a) and 8(b), PSNR maps are plotted for standard FFDM and SEDM, respectively. As discussed previously, the PSNRs can be used to quantify and compare the low-contrast performance for different imaging techniques without the presence of a low-contrast object. Both the air regions and the periphery of the breast phantom were excluded (by setting the threshold at four times the minimum signal level in the FFDM image) in the PSNR measurements. The PSNRs tended to be lower in denser tissue areas but remained higher in less dense tissue areas and at the periphery for the FFDM image. However, the PSNRs appeared to be largely uniform in the SEDM image within the entire compressed breast area. The overall PSNRs were measured to be 79.6 for the FFDM and 107.6 for the SEDM imaging on average in the entire compressed breast phantom area. The standard deviations of PSNRs were estimated to be 10.3 and 7.9 for FFDM and SEDM imaging, respectively. The PSNRIFs for the exposure equalization are shown in Fig. 8(c). We found that the PSNRIF tended to be higher to compensate for the heavy attenuation in denser tissue areas. We found that the highest achievable PSNRIF was 1.89 for the anthropomorphic breast phantom used in this study, and the average PSNR was found to improve by 35% over the entire breast phantom.

Figures 9(a) and 9(b) show the spatial distribution of the pixelated glandular dose $D_g(i, j)$ estimated in the breast phantom for FFDM and SEDM, respectively. At the breast periphery, the glandular dose increased as the breast thickness decreased for FFDM. However, the glandular dose decreased at the breast periphery for SEDM as a result of exposure equalization, which had led to reduced exposure in the breast periphery region. The average glandular dose was reduced

FIG. 8. (a) PSNRs measured in the FFDM image; (b) PSNRs measured in the SEDM image; and (c) PSNR improvement factors (PSNRIFs) measured as a ratio of PSNR for the SEDM image over that for the FFDM image.

by $\sim 34\%$, from 323.8 mrad for FFDM to 212.5 mrad for SEDM. With the breast periphery excluded, the average glandular dose was increased by $\sim 12\%$, from 247.9 mrad for FFDM to 277.6 mrad for SEDM. The spatial variation of RD_g is shown in Fig. 9(c). The highest value of RD_g was found to be 1.67, in the densest breast tissue area. RD_g was reduced substantially at the breast periphery, as expected because of the reduction of exposure in the breast periphery region.

IV. DISCUSSION AND CONCLUSIONS

In this study, we used standard FFDM images of an anthropomorphic breast phantom acquired with an anti-

scatter grid at various exposure levels to simulate SEDM imaging. We investigated the low-contrast performance of SEDM imaging and the performance improvement by measuring and comparing the PSNRs with those of standard FFDM. We found that the PSNRs appeared to be largely uniform with exposure equalization, and the standard deviations of PSNRs were estimated to be 10.3 and 7.9 for FFDM and SEDM imaging, respectively. The overall PSNRs were measured to be 79.6 for FFDM and 107.6 for SEDM on average in the compressed breast phantom area, resulting in an average improvement of the PSNR by $\sim 35\%$ with exposure equalization. The highest achievable PSNRIF was found to be 1.89 for the anthropomorphic

FIG. 9. Spatial distribution of glandular dose estimated in the entire breast phantom for (a) FFDM and (b) SEDM. (c) Spatial distribution of RD_g estimated in the entire breast phantom.

breast phantom used in this study. The effect of exposure equalization on radiation dose was estimated by computing the average glandular dose in the entire breast phantom. Exposure equalization resulted in an average glandular dose reduction of $\sim 34\%$ compared with standard FFDM. It should be noted that the results from this study apply only to the specific detector/grid combination used for standard FFDM, with the assumption that the scatter properties of FFDM imaging are equivalent to those of slot-scan imaging with a 15-mm-wide slot. Compared with the earlier beam compensator approach, an additional advantage of the SEDM method is that scattered radiation near the breast periphery can be reduced substantially as a result of reduced exposure at the breast periphery without adding a scatter medium (beam compensator) to the breast. It should be noted that the effect of exposure equalization on scatter properties and low-contrast performance may be more limited in breast imaging than in chest imaging. This is due mainly to the differences in anatomical structure between the two imaging objects. The image signal variation was observed to be $\sim 2\text{--}3\text{:}1$ in mammography, relatively lower than that in chest radiography (as high as $100\text{:}1$).⁴⁵

We have introduced a simple method that uses only a limited number of FFDM images acquired at various exposure levels to construct an SEDM image. However, this method has some limitations. First, this method applies only to the specific study conducted here. The PSNRs and their improvement can be correctly quantified for the exposure equalization along with the noise model and correction method used in this study, but this method may not work very well with more sophisticated studies, such as image perception, where any changes in noise properties may affect the results of reading studies. Second, the change of scattering properties due to the exposure equalization and its effects on PSNRs could not be simulated with this method. In fact, the overall change in PSNR for an ROI is a combined effect of noise and scattered radiation (from surrounding tissues). For instance, in SEDM, the exposure to the central dense tissue mass should increase while the exposure to the surrounding fatty tissue area should decrease in the case of a small, dense tissue mass surrounded by a large fatty tissue area. Thus, the noises in the dense tissue mass area will be lower as a result of increased exposure to this area and reduced scattering from the surrounding fatty tissue. In contrast, in the case of a small fatty tissue area surrounded by a dense tissue mass, the exposure to the central fatty tissue should decrease while the exposure to the surrounding dense tissue mass should increase. Thus, the noises in the central fatty tissue area will be higher as a result of reduced exposure to this area and increased scattering from the surrounding dense tissue. However, the SPRs are already low in standard mammography (~ 0.22 on average with an anti-scatter grid), and the effects of scattering properties on PSNRs due to the exposure equalization may be limited. Such effects can only be investigated when an actual SEDM system is implemented.

The anthropomorphic breast phantom uses a single breast-equivalent material (BR-12) to mimic the attenuation pattern via thickness variations. The average glandular dose for this

breast model was estimated on the basis of the summation of the pixelated glandular doses derived by the following assumptions: first, the pixelated breast thickness was simulated from the primary intensity image of the breast phantom; second, the corresponding pixelated breast thickness image was used to estimate the pixelated $D_{gN}(i, j)$ on the basis of the published data.⁴⁴ The result showed that the estimated average glandular dose for FFDM was 247.9 mrad when the breast periphery was excluded. The average glandular dose can also be calculated directly by multiplying X_{ESE} by D_{gN} ; the result was 241.4 mrad. This indicates that using the proposed method to estimate the average glandular dose is probably acceptable. However, the published data were estimated on the basis of a phantom model with a uniform mixture of glandular tissue and adipose tissue for a given breast thickness. As a result, this difference affects not only the scattering pattern in the breast phantom but also the pattern of pixelated glandular tissue dose. The average glandular dose can be accurately estimated only with Monte Carlo simulation, and such a study is beyond the scope of this work.

Another concern related to the development of an SEDM system is the x-ray tube heat loading when slot-scan geometry is used. The x-ray tube current and generator output are usually limited in a conventional SF or digital mammography system because of the low-energy nature of mammography x-rays and the requirement for small focal spot size. The x-ray usage efficiency in slot-scan imaging is generally low (only a few percent that of conventional full-field mammography), depending on the slot width used. Therefore, a high-current x-ray tube and high-power generator are essential for slot-scan digital mammography. A tungsten target-based x-ray tube capable of high current output with a specially designed high-power generator has been successfully developed and used in a commercial slot-scan digital mammography (SSDM) system (SenoScan, Fischer Imaging Corporation, Denver, CO).^{46–48} However, the requirement for x-ray tube current and generator output is even higher for SEDM imaging than for slot-scan imaging because higher mAs is required when the fan beam exposes an area containing dense tissue structures.³⁸

Several factors must be considered in selecting x-ray tube and generator for construction of the proposed SEDM system. In general, the generator power rating is limited by the x-ray tube in an x-ray machine. The slot width and exposure time are key factors to determine the maximum allowable tube current in SEDM imaging. Narrower slot width results in better scatter rejection and improved low-contrast performance.⁴⁰ Shorter exposure time helps reduce comfortlessness to the patient. However, they both require a higher tube current to maintain sufficient x-ray exposure in order to preserve the image quality. The maximum tube current of the commercial SSDM system is 220 mA and the scanning time is approximately 6 S, which is considered adequate. The SSDM system has a slot width of approximately 10 mm, resulting in an effective exposure time of 0.2 S and an effective exposure corresponding to 44 mAs at tube current of 220 mA for a scanning distance of 300 mm. The high performance of scatter rejection is achieved with low x-ray usage efficiency of

approximately 3.3% (10/300 mm) only of generated x-rays. A slot width of 15 mm has demonstrated a comparable effectiveness of scatter rejection when compared with anti-scatter grid in this study. Hence the effective exposure time and the mAs corresponding to the effective exposure can be increased to 0.3 S and 66 mAs, respectively, with a slot width of 15 mm. We have found that up to three times exposure higher than that of AEC timed was required in dense tissue regions for the anthropomorphic breast phantom used in this study. Therefore the effective exposure of 66 mAs may not be high enough for the proposed SEDM system. An even more powerful x-ray tube with higher tube current and heat loading capacity is needed in construction of an SEDM system.

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