Breast Imaging Using an Amorphous Silicon-Based Full-Field Digital Mammographic System: Stability of a Clinical Prototype

Srinivasan Vedantham, Andrew Karellas, Sankararaman Suryanarayanan, Carl J. D'Orsi, and R. Edward Hendrick

An amorphous silicon-based full-breast imager for digital mammography was evaluated for detector stability over a period of 1 year. This imager uses a structured CsI:Tl scintillator coupled to an amorphous silicon layer with a 100-micron pixel pitch and read out by special purpose electronics. The stability of the system was characterized using the following quantifiable metrics: conversion factor (mean number of electrons generated per incident x-ray), presampling modulation transfer function (MTF), detector linearity and sensitivity, detector signal-to-noise ratio (SNR), and American College of Radiology (ACR) accreditation phantom scores. Qualitative metrics such as flat field uniformity, geometric distortion, and Society of Motion Picture and Television Engineers (SMPTE) test pattern image quality were also used to study the stability of the system. Observations made over this 1-year period indicated that the maximum variation from the average of the measurements were less than 0.5% for conversion factor, 3% for presampling MTF over all spatial frequencies, 5% for signal response, linearity and sensitivity, 12% for SNR over seven locations for all 3 target-filter combinations, and 0% for ACR accreditation phantom scores. ACR mammographic accreditation phantom images indicated the ability to resolve 5 fibers, 4 speck groups, and 5 masses at a mean glandular dose of 1.23 mGy. The SMPTE pattern image quality test for the display monitors used for image viewing indicated ability to discern all contrast steps and ability to distinguish line-pair images at the center and corners of the image. No bleeding effects were observed in the image. Flat field uniformity for all 3 target-filter combinations displayed no artifacts such as gridlines, bad detector rows or columns, horizontal or vertical streaks, or bad pixels. Wire mesh screen images indicated uniform resolution and no geometric distortion.

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SCREEN-FILM TECHNOLOGY currently is the standard in full-breast imaging for the detection of nonpalpable breast cancer. Advancements in digital technology in the past decade have made it feasible to obtain high-quality images using digital detectors, and earlier this year the US Food and Drug Administration approved the first full-breast digital imaging system. Screen-film mammography is very effective for screening, but it has well-known limitations with regard to dy-

namic range, contrast, and lack of convenient options for postprocessing of images. Digital detectors have the theoretical capability to overcome some of these fundamental limitations. Digital detectors also show the potential to provide high detection efficiency, high dynamic range, capability of contrast enhancement¹ and postprocessing capabilities including computer-aided diagnosis.²⁻⁷ In addition, with the advent of digital detectors it is feasible to perform novel imaging techniques like tomosynthesis,⁸⁻¹¹ digital subtraction,¹² and dual-energy mammography.^{13,14}

Different modes of electronic detection like charge-coupled devices¹⁵⁻¹⁹ (CCDs) and image intensifiers have been used in the past. A detailed study on a CCD-based whole-breast digital mammography unit with emphasis on establishing performance standards based on long-term stability also has been reported.²⁰ Improvements in thin-film transistor (TFT) technology and the ability to manufacture large flat panel semiconductor monolithic arrays have resulted in increased interest in utilizing these arrays for full-breast digital mammography. Currently, amorphous silicon²¹⁻²⁷ and amorphous selenium^{28,29} semiconductors are sub-

From the Department of Radiology, UMass Memorial Health Care, University of Massachusetts Medical School, Worcester, MA, and the Radiology Department, University of Colorado Health Science Center, Denver, CO. (Presently at Lynn Sage Comprehensive Breast Center, Northwestern University Medical School, Chicago, IL.)

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Address reprint requests to Andrew Karellas, PhD, Professor of Radiology, Department of Radiology, Room S2-836, University of Massachusetts Medical School, UMass Memorial Health Care, 55 Lake Ave North, Worcester, MA 01655.

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jects of active research for imaging applications. Amorphous silicon-based detectors use a scintillator for conversion of x-ray photons to light photons; the generated light photons subsequently are detected by the amorphous silicon photodiode unlike the direct conversion photoconductor used by amorphous selenium-based detectors.

This study characterizes the stability of an amorphous silicon-based clinical prototype flat panel imager (GE Medical Systems, Milwaukee, WI), which underwent technical and clinical evaluation at the University of Massachusetts Medical School and the University of Colorado Health Sciences Center. Although this study characterizes the stability of a clinical feasibility system, the stability of a clinical system, which has a larger pixel matrix (1920×2304), is the subject of an ongoing study.

MATERIALS AND METHODS

The full-breast digital mammography imager used in this study consists of a thallium-doped CsI:Tl scintillator and an amorphous silicon photodiode array and incorporates special-purpose readout electronics. Light created from the interaction of x-ray photons in the scintillator travels down the columnar crystalline structure of the scintillator, which is in contact with a 2-dimensional array of amorphous silicon photodiodes and thin-film transistors. Light exiting from the scintillator is detected by the monolithic thin film flat panel array, which consists of a matrix of 1800×2304 pixels that are $100 \mu\text{m}$ in pitch. The specifications of the flat panel imager are presented in Table 1. Each pixel in the array is an individually addressable light detector. The electrical signals of all pixels are individually read out and digitized to 16-bit digital values in 300 ms by special-purpose low-noise electronics,³⁰ which are located inside the image receptor assembly. The schematic of the detector is shown in Fig 1. The imager is integrated into a prototype digital mammography system based on a multipulse high-frequency x-ray generator (Senographe DMR, GE Medical Systems). This system uses an x-ray tube with a selectable dual track target, either molybdenum (*Mo*) or rhodium (*Rh*) and selectable filtration of *Mo* or *Rh*. The stability of the system was quantified in terms of the following metrics: conversion factor (number of electrons detected per incident x-ray), modulation transfer function (MTF), detector signal-to-noise ratio (SNR) measurement, flat field uniformity, and ACR accreditation phantom scores.

Table 1. Specifications of the Amorphous Silicon-Based Flat Panel Full-Breast Digital Mammographic Imager

Image area	18 cm $\times$ 23 cm
Image matrix	1800 $\times$ 2304
Scintillator	CsI: Tl
Pixel pitch	100 microns

Fig 1. Schematic of the amorphous silicon-based flat panel full-breast digital mammographic imager.

Conversion Factor

The conversion factor is defined as the mean number of electrons detected per incident x-ray. The number of incident photons was estimated using an x-ray spectral model built into the system as part of the quality control tests and provided by the manufacturer. Because this x-ray spectral model addresses only the 30 kVp, *Rh* target and *Rh* filter spectrum, measurements were performed with an exposure of 30 kVp, *Rh* target and *Rh* filter at 20 mAs as per the manufacturer recommended procedure. The mean number of generated electrons was computed using the algorithm provided with the system, which takes into account the mean number of digital exposure units of the central 256×256 pixels of the full-breast detector and the gain of the amplifier. This measurement was performed over a period of 1 year on a monthly basis to assess variations in conversion factor.

Presampling MTF

The presampling MTF of the system was characterized using the edge response function (ERF) method. The ERF was obtained using a 0.125-mm thick tungsten bar placed at a slight angle ($<5^\circ$) as shown in Fig 2. Because each pixel represents a sample of the ERF at a distance equal to the center of the pixel to the edge, the oversampled ERF was obtained by plotting the pixel intensity from the edge. This technique is similar to the methodology used by Fujita et al³¹ to obtain the oversampled LSF. The line-spread function (LSF) was derived from the ERF by differentiation. The modulus of the Fourier transform of the LSF provided the presampling MTF of the system. The ERF was obtained by exposing the tungsten bar at 30 kVp, *Rh* target, *Rh* filter, and 20 mAs as per manufacturer's recommendations. This measurement was performed over a period of 1 year on a monthly basis to assess variations in MTF. An alternate technique using a 10-micron slit to obtain the LSF also was used to confirm the results obtained.²⁶

Detector Linearity and Sensitivity

The linear response was measured at 28 kVp, *Mo* target, *Mo* filter, and added filtration of 4.5 cm of Lucite, by averaging the pixel intensities over a 256×256 area located at the center of the image at various exposure levels. The sensitivity of the

Fig 2. Illustration of the angular placement of the tungsten bar to obtain the edge response function (ERF) for modulation transfer function (MTF) measurements.

system was determined by evaluating the slope of the linear response curve.²⁶

Detector Signal-to-Noise Ratio

The detector SNR was measured at 7 locations across the imager for all 3 target-filter combinations. A 1-inch thick uniform sheet of Lucite covering the entire detector was placed on the breast support plate. The exposure techniques for the three target-filter combinations are shown in Table 2. Exit exposures (exposure on to the breast support plate) were measured with the 1-inch-thick Lucite sheet elevated above the ion chamber with a 10-cm air gap between the 2 objects similar to the technique used by Kimme-Smith et al.²⁰ The acquired images were dark-image subtracted and flat-field corrected²⁵ and viewed at the image-review station (Sun Microsystems, Palo Alto, CA). The image-review station is equipped with dual CRT displays driven by special purpose display-driver boards (Dome Imaging Systems, Inc, Waltham, MA). The mean pixel intensity, μ (digital values) and standard deviation, σ , were measured using a 539 mm² region of interest (ROI) at 7 locations of the image as shown in Fig 3. The SNR was

Fig 3. Illustration of the location of the ROI for SNR measurements.

obtained by dividing the mean pixel intensity by the standard deviation (μ/σ).

ACR Accreditation Phantom Scores

An American College of Radiology (ACR) Mammography accreditation phantom (Model: 18-220, Nuclear Associates, Cleveland, OH) was imaged using the following technique factors: *Mo/Mo*, 25 kVp, 100 mAs, large focal spot, 660 mm source-to-image distance (SID), with antiscatter grid. This technique corresponds to a mean glandular dose of 1.23 mGy. The images were viewed at the image-review station. Optimum windowing and leveling, and magnification of the image up to a factor of 1.7 was utilized for scoring the phantom. Phantom scoring was performed according to ACR guidelines.³² Then the images were printed onto a dry laser compatible film (DryView Laser Imaging Film, Eastman Kodak Co, Rochester, NY) using a dry-film laser printer (DryView 8600, Eastman Kodak Co) and scored according to ACR guidelines.³²

Qualitative Measures

In addition to these quantitative measurements, qualitative measurements to characterize the stability of the system to provide consistent image quality also were performed. These studies included flat-field uniformity, geometric distortion, and Society of Motion Picture and Television Engineers, White Plains, NY (SMPTE) test pattern image quality for the display monitors used for patient image viewing.

The flat-field uniformity of the detector was studied by imaging a 1-inch thick uniform sheet of Lucite covering the entire detector placed on the breast support plate, using the technique factors listed in Table 2 for all 3 target-filter combinations. The images were studied at the image-review station for artifacts such as gridlines, bad detector rows or columns, horizontal or vertical streaks, and bad pixels.

Table 2. Exposure Techniques Used for All 3 Target-Filter Combinations for SNR Measurements and Flat-Field Uniformity Measurements

Target-Filter	Tube Voltage (kVp)	Tube Current $\times$ Time (mAs)	Exit Exposure (mR)
<i>Mo-Mo</i>	25	110	63.9
<i>Mo-Rh</i>	28	80	72.0
<i>Rh-Rh</i>	32	40	67.1

NOTE. The exit exposure represents the exposure on to the breast support plate after transmitting through 1-inch of Lucite.

Fig 4. The SMPTE test pattern with the 0% to 5% and 95% to 100% contrast boxes labeled.

Geometric distortion was evaluated by imaging a wire mesh screen at 25 kVp, *Mo/Mo*, 50 mAs, 660 mm SID, and with the antiscatter grid. The image was visualized at the image-review station and evaluated for resolution uniformity and distortion.

The SMPTE pattern image quality was used to assess the stability of the CRT displays used for patient image viewing at the image-review station. The photometer test tool provided by Dome was utilized for generating the SMPTE pattern. An image of the SMPTE test pattern is shown in Fig 4 with the 0% to 5% and 95% to 100% contrast boxes labeled. An extensive description of the use of SMPTE test pattern images for evaluating soft copy devices was given by Nawfel et al.³³ The SMPTE pattern image was studied for the following features: ability to discern 0% to 5% contrast box and 95% to 100% contrast box, ability to distinguish line-pair images at the center and corners of the SMPTE pattern, and presence of any bleeding effects.

RESULTS AND DISCUSSION

The conversion factor measured over a period of 1 year as shown in Fig 5 was stable and indicated very little variability. The maximum variation from the mean was less than 0.5%. The mean presampling MTF measured is shown in Fig 6. The error bars represent the maximum variations observed over the period of 1 year at discrete spatial frequencies. The presampling MTF measured over a period of 1 year at 1, 2, 3, 4, and 5 cycles per

millimeter are shown in Fig 7. The maximum variation from the mean at these discrete spatial frequencies was less than 3% over a period of one year. This variation can be attributed partially to experimental error.

Fig 5. Conversion factor measurements over a period of 1 year performed on a monthly basis. Maximum variation was less than 0.5% during this period.

Fig 6. Average of the presampling MTF measurements made over a period of 1 year on a monthly basis. The error bars represent the maximum variations observed over this duration at discrete spatial frequencies.

The measured detector linearity is shown in Fig 8. The error bars represent the maximum deviations from the average of the mean pixel intensity (digital units) over a 256×256 ROI observed over a period of 1 year. From the linearity measure-

ments the mean sensitivity of the system was found to be 16.3 digital units per milliroentgen per pixel. Maximum variation in linearity and sensitivity was less than 5% over this period.

The signal, defined as the mean of each of the 539 mm^2 ROI as shown in Fig 3, varied by less than 5% (4% for *Mo-Mo*, 4% for *Mo-Rh*, and 2% for *Rh-Rh*) over the 7 locations for all 3 target-filter combinations. The coefficient of variation defined as the ratio of the standard deviation to the mean ($\sigma/\mu \times 100\%$),²⁰ in SNR measurements over the 7 locations of the detector was less than 7% for the *Mo-Mo*, less than 9% for the *Mo-Rh*, and less than 4% for the *Rh-Rh* target-filter combinations. As an example, one such SNR measurement is shown in Fig 9. The SNR measurements seem to indicate a trend in which the SNR decreases from the edge of the detector proximal to the chest wall (ROI locations 1, 2, and 3) toward the edge of the detector distal to the chest wall (ROI locations 5, 6, and 7). To verify if this trend was caused by the heel effect, the signal amplitudes of the ROIs 2, 4, and 6 were analyzed for each of the 3 target-filter

Fig 7. The presampling MTF measured over a period of 1 year at discrete spatial frequencies of 1, 2, 3, 4, and 5 cycles per millimeter (cy/mm). The maximum variation from the mean at these discrete spatial frequencies was less than 3% over this period.

Fig 8. The measured detector linearity. The error bars represent maximum deviations from the average of the mean pixel intensity (digital units) of a 256 pixels $\times$ 256 pixels ROI observed over a period of 1 year.

combinations. The signal amplitudes after flat-field correction from these 3 ROIs did not indicate any fall-off trend. This suggests that the flat-field correction technique was adequate to suppress the heel effect. The heel effect would result in higher exposure at the edge proximal to the chest wall compared with the edge distal to the chest wall. This would result in higher signal amplitudes at the edge proximal to the chest wall compared with the edge distal to the chest wall. Flat-field correction techniques would compensate this trend resulting in an almost uniform signal throughout the entire field. However, under quantum-limited operating conditions, a difference in exposure levels between the edge proximal to the chest wall and the edge distal to the chest wall would lead to a trend in the SNR proportional to the square root of the difference in the number of x-ray photons. Hence, the ratio of the SNR obtained from ROIs 2 and 6 were compared with the square root of the ratio of the exposure levels at ROIs 2 and 6. This ratio indicated that the measured SNR fall-off trend was within 5% of the expected SNR fall-off trend. This suggests that the flat-field correction technique was adequate in flattening the field (signal) and the difference in the exposure levels between the edge proximal to the chest wall and the edge distal to the chest wall is a major contributor to the observed SNR fall-off trend. The mean SNR over the 7 locations was 121 for the *Mo-Mo*, 138 for the

Mo-Rh, and 130 for the *Rh-Rh*. The increases (%) in mean SNR with the *Mo-Rh* and *Rh-Rh* target-filter combinations compared with the *Mo-Mo* target-filter combination were 14% and 7.5%. The improvement in the mean SNR for the *Mo-Rh* and *Rh-Rh* target-filter combinations are attributed partially to the higher exit exposures (12.6% higher for the *Mo-Rh* and 5% higher for the *Rh-Rh* target-filter combinations compared with the *Mo-Mo* target-filter combination).

Images of the ACR accreditation phantom scored both with the hard copy (printed film) and the soft copy (CRT display) showed the ability to resolve 5 fibers, 4 speck groups, and 5 masses at breast doses comparable to that of screen-film phantom images. These images did not show any variability over the 1-year period. An ACR accreditation phantom image acquired at 25 kVp, 100 mAs, *Mo-Mo* target-filter combination, 660 mm SID, with antiscatter grid corresponding to a mean glandular dose of 1.23 mGy is shown in Fig 10. The process of transferring the image on to a paper print resulted in a slight degradation, but the hard copy (printed film) and soft copy (CRT display) images indicated the capability of resolving 5 fibers, 4 speck groups, and 5 masses.

The SMPTE pattern image quality test indicated ability to discern all contrast steps, 0% to 5% contrast box, 95% to 100% contrast box, and ability to distinguish line-pair images at the center and corners of the SMPTE pattern image. No

Fig 9. SNR measurements over the 7 locations of the detector as shown in Fig 3 for all 3 target-filter combinations. Observed variation was less than 12% from the central ROI 4.

Fig 10. An ACR accreditation phantom image acquired at 25 kVp, 100 mAs, *Mo/Mo* target/filter combination, 660 mm SID, with antiscatter grid corresponding to a mean glandular dose of 1.23 mGy. The process of transferring the image onto a paper print resulted in a slight degradation, but the hard copy (printed film) and soft copy (CRT display) images indicated the capability of resolving 5 fibers, 4 speck groups, and 5 masses.

bleeding effects were observed in the SMPTE pattern image. The flat-field uniformity study conducted for all 3 target-filter combinations did not display artifacts like gridlines, horizontal or vertical streaks, or bad pixels. Wire mesh screen images indicated uniform resolution and no geometric distortion.

The most common and regularly used measure of image quality has been the scoring of ACR mammographic accreditation phantom. Also, a compelling argument has been presented with respect to the requirement of improved performance to justify the increased costs.²⁰ Based on these factors and the improved physical characteristics observed with such digital mammographic systems,²⁶ a logical requirement for these full-field

digital mammographic systems might be the need to perform better than screen-film systems. We were able to detect regularly 4 speck groups and, probably more significantly, 5 masses, which denotes performance that exceeds that expected of screen-film systems. A practical quality assurance procedure should probably include the scoring of the ACR accreditation phantom on a weekly basis. In addition, because the ability to achieve consistent imaging performance would depend on the flat-field uniformity, performing gain, offset, and flat field calibrations on a weekly basis might be appropriate.

The full-breast digital mammographic imager did not exhibit any appreciable artifact or structured noise. The flat panel imager showed good

stability as observed by the conversion factor measurements. Although variations in detector signal response, linearity, sensitivity, and SNR were observed, interim results from a multicenter clinical trial do not show a significant difference between film-screen and full-field digital mammography in sensitivity for breast cancer detection.³⁴ The flat panel imager also showed a remarkable capability to image fibrils, speck groups, and masses in the ACR phantom. Clinical images show equally favorable results in visualizing soft tissue anatomy and calcifications.³⁴⁻³⁶ Although the phantom and clinical images are very encouraging, the clinical efficacy in terms of sensitivity and specificity is the subject of a different investigation, which is currently in progress.³⁴

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