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Full breast digital mammography with an amorphous silicon-based flat panel detector: Physical characteristics of a clinical prototype

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Abstract

The physical characteristics of a clinical prototype amorphous silicon-based flat panel imager for full-breast digital mammography have been investigated. The imager employs a thin thallium doped CsI scintillator on an amorphous silicon matrix of detector elements with a pixel pitch of 100 μm . Objective criteria such as modulation transfer function (MTF), noise power spectrum, detective quantum efficiency (DQE), and noise equivalent quanta were employed for this evaluation. The presampling MTF was found to be 0.73, 0.42, and 0.28 at 2, 4, and 5 cycles/mm, respectively. The measured DQE of the current prototype utilizing a 28 kVp, Mo–Mo spectrum beam hardened with 4.5 cm Lucite is ~55% at close to zero spatial frequency at an exposure of 32.8 mR, and decreases to ~40% at a low exposure of 1.3 mR. Detector element nonuniformity and electronic gain variations were not significant after appropriate calibration and software corrections. The response of the imager was linear and did not exhibit signal saturation under tested exposure conditions.

Keywords

breast imaging; digital mammography; physics; image quality; detective quantum efficiency (DQE)

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I. INTRODUCTION

The physical aspects of mammography have been the subject of many investigations which have addressed basic imaging characteristics such as x-ray scatter,^{1–4} x-ray tube focal spot effects,⁵ and x-ray spectra.^{6,7} This knowledge has served as the basis for many technical improvements and regulatory standards of performance.⁸

Though film-screen mammography is currently the standard in breast imaging, it has well-known limitations with regard to dynamic range, contrast, and lack of convenient options for postprocessing of images. It is apparent that electronic detection has the theoretical capability of overcoming certain fundamental limitations of film-screen systems. The potential advantages of electronic detection include high detection efficiency, high dynamic range, capability of contrast enhancement,⁹ and postprocessing capabilities including computer-aided diagnosis.^{10–15} Further, direct electronic acquisition enables the exploration of novel imaging techniques such as tomosynthesis,^{16,17} dual-energy mammography,^{18,19} and digital subtraction imaging.²⁰ In the past, investigators have used different modes of electronic detection technology to gain insight into electronic mammography, commonly referred to as digital mammography.²¹ Early evaluations have used image intensifiers and subsequently slot-scanned systems^{22,23} with charge-coupled devices (CCDs) and CCDs with fiberoptic tapers.²⁴ Development of an electronic detector to cover the entire breast presents a formidable technical challenge. Currently, digital mammography is limited to small field devices for stereotactic localization, core biopsy, and spot compression views.^{24,25} It is now feasible to manufacture large flat panel monolithic arrays of amorphous silicon photodiodes coupled to thin-film transistors on a glass substrate. These arrays utilize a scintillator as the primary detection layer to convert x rays to light, which is subsequently detected by the photo-sensing silicon elements. Several studies characterizing amorphous silicon^{26–30} and amorphous selenium^{31,32} based imagers for chest radiography and other applications have been reported in the recent past. However, detailed experimental characterization of amorphous silicon based flat panel imagers under realistic mammographic conditions have not been reported in the past.

This study characterizes the image quality parameters of an amorphous silicon-based clinical prototype flat panel imager (GE Medical Systems, Milwaukee, WI) presently undergoing technical and clinical evaluation at the University of Massachusetts Medical School and the University of Colorado Health Sciences Center.

II. METHODS AND MATERIALS

The full-breast digital mammography imager characterized in this study is composed of a thallium-doped CsI 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 two-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 detector elements that are $100 \mu\text{m}$ in pitch. The specifications of the mammographic flat panel imager are presented in Table I.

Each detector element (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 electronics³³ 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, Milwaukee, WI). This system uses a selectable dual track target, either molybdenum (Mo) or rhodium (Rh) with selectable filtration of Mo or Rh. All measurements were performed at 28 kVp with a Mo/Mo target/filter combination. This particular technique was chosen as it was found to be the median exposure technique used in a random sample of 100 breast exams from a population of 1400 patients performed with this flat panel imager.

A. Presampling modulation transfer function measurement

The presampling modulation transfer function (MTF) was measured according to the technique described by Fujita *et al.*³⁴ The experimental procedure for measuring the same has also been described in detail by Dobbins *et al.*³⁵ The effects of undersampling have also been described in detail by Dobbins.³⁶ The experimental setup is shown in Fig. 2. An image of a 10-mm-long, 10 μm ($\pm 1 \mu\text{m}$) slit made of 1.5-mm-thick tantalum placed at a slight angle (less than 4°) to the anode–cathode axis at the center of the detector was obtained. The area around the slit was covered with Pb (0.5 cm thick). The slit was placed about 5.5 mm (due to thickness of the breast support plate and the slit housing) from the surface of the imager. Since the magnification of the slit was about 1.0083, there was no appreciable spreading of the line spread function (LSF) due to focal spot blurring. The exposure technique was adjusted to ensure that the tails of the dark image subtracted LSF obtained had no significant electronic noise. The appropriate technique found to be 28 kVp, 160 mAs was used. The source-to-image distance was maintained at 660 mm during the study. The image of the slit was obtained without the antiscatter grid in place. The slit image obtained was corrected for variations along the edge of the slit. This was accomplished by normalizing the signal values along the horizontal direction (perpendicular to the anode–cathode axis) by dividing each pixel value by the sum of the pixel values in that particular row as illustrated in Fig. 3. This normalization method assumes that the slit width is approximately constant over the length used for obtaining the finely sampled LSF and that the signal spreading is approximately equal along each line of data. The validity of these assumptions was verified by calculating the MTF from several locations along the central region of the slit, and the MTF was found to vary by less than 1%. Before performing this normalization care was taken to avoid loss of information due to truncation by converting the pixel intensity values to 32 bit floating point numbers. The pixel amplitudes along the column or vertical direction (along the anode–cathode axis) were plotted as shown in Fig. 4. This provided the adequate number of individual LSFs needed to obtain a finely sampled LSF. Since each pixel represented a sample of the LSF at a distance equal to the distance between the center of the slit and the pixel center, the finely sampled LSF was obtained by plotting the pixel intensity from the center of the slit. The finely sampled LSF was synthesized by using 34 individual LSFs and normalized to a peak value of one (Fig. 5). The Fourier transform (FT) of the finely sampled LSF was performed and the resultant FT was deconvolved of the finite dimension of the slit by dividing the resultant FT by a sinc function in the frequency domain to provide the

presampling MTF. The presampling MTF was measured along both the horizontal (perpendicular to the anode–cathode axis) and vertical (along the direction of the anode–cathode axis) directions.

B. Noise power spectrum measurement

There are many inherent difficulties in measuring the noise power spectrum (NPS) of digital systems.^{35–38} Computing the two-dimensional (2D) NPS is important to study the presence or absence of any off-axis noise peaks. Since the computation time of computers is no longer a constraint,³⁵ computing the entire 2D NPS and estimating the one-dimensional (1D) NPS from the 2D NPS was used. The 1D NPS was estimated from the 2D NPS using the technique described by Dobbins *et al.*³⁵ This technique utilizes a thick cut parallel to and immediately adjacent to the axes for estimating the 1D NPS. We used the data in a thick slice comprised of eight lines on either side of both the axes (excluding the axes). For each data value at (u,v) in this thick slice, the frequency value was computed as $\sqrt{u^2+v^2}$ for the 1D NPS estimate. The assumptions for utilizing this technique for estimating the 1D NPS are that the 2D NPS exhibit moderate radial symmetry and that the noise data are nominally uniform within the small annuli of spatial frequencies used for regrouping the noise data.

The next major difficulty was to determine the finite window of the noise data required to provide adequate resolution for proper representation of the NPS without the finite window overtly affecting the NPS estimate. Since the measured NPS is produced by convolving the “true” NPS with the sinc² function in the frequency domain, due to the finite window of the noise data, the choice of region-of-interest (ROI) size has to be considered carefully. We estimated the NPS using ROI sizes of 512×512, 256×256, 128×128, and 64×64, and determined the 256×256 ROI to be the smallest ROI required for proper representation of the NPS with minimum spectral distortion (spectral deviation between 512×512 ROI and 256×256 ROI was less than 5% over the entire frequency range and the spectral deviation increased with smaller ROI sizes). Hence, the 256×256 ROI was utilized for NPS estimations in the entire study.

The other difficulty was to determine the number of NPS realizations needed to be averaged in order to obtain a smooth and accurate curve depicting the noise spectrum. Ideally, we would need a large number of NPS realizations so that they can be averaged to obtain a smooth spectrum. We considered 10, 15, 20, 30, and 50 NPS realizations and found that the ensemble average of 15 NPS realizations taken from the same location through 15 images was sufficient to accurately characterize the NPS of the system. We were able to achieve a smooth spectrum by averaging eight lines of data on either side of the axes.

Problems associated with background trends such as from the heel effect can corrupt the noise spectrum and provide artificially inflated values^{35,38} along the axes. However, techniques for suppression of such background trends have been described by various authors.^{35,38} We surface (ramp) fitted each ROI and subtracted these background trends. Though this method was successful in suppressing these background trends, it did not completely eliminate them. Hence, we avoided using data values directly on the axes, as

they were not representative in amplitude of the rest of the 2D NPS in the vicinity of the axes.

In order to measure the noise power spectra of the detector the detector has to be linear and shift invariant.³⁹ The linear response and sensitivity of the system was measured by averaging the pixel intensity over a 256×256 ROI centered at the 4 cm from the chest wall edge of the detector at various exposure levels. All images for the noise power spectral estimate used for calculation of detective quantum efficiency (DQE) were dark subtracted [Eq. (1)] and flat field corrected [Eq. (2)] resulting in a nominally uniform image,

$$\text{dark subtracted}_i(x, y) = \text{flood}_i(x, y) - \text{dark}_i(x, y), \quad (1)$$

$$\begin{aligned} \text{flat field}_i(x, y) = & \frac{\text{dark subtracted}_i(x, y)}{(1/n) \sum_{i=1}^n \text{dark subtracted}_i(x, y)} \\ & \times \frac{1}{m^2} \sum_{v=1}^m \sum_{u=1}^m \left[\frac{1}{n} \sum_{i=1}^n \text{dark subtracted}_i(x, y) \right], \quad (2) \end{aligned}$$

where $\text{flood}_i(x, y)$ and $\text{dark}_i(x, y)$ represent the flood and dark ROIs, respectively;

$(1/n) \sum_{i=1}^n \text{dark subtracted}_i(x, y)$ is the average of the dark subtracted ROIs; $1/m^2 \sum_{v=1}^m \sum_{u=1}^m [(1/n) \sum_{i=1}^n \text{dark subtracted}_i \times (x, y)]$, is the mean of the average of the dark subtracted ROIs; and, in our case, $m=256$ and $n=15$. The ROIs (256×256) used for the NPS analysis were taken from the same location (centered at 4 cm from the chest wall edge of the detector) from multiple (15) images. Though the detector might not to be completely shift invariant, the process of flat field correcting and using the same ROI from multiple images for NPS analysis allows for the reasonable assumption of the “shift-invariant” property of the system.

The noise power spectra were determined at four exposure levels and were obtained with 4.5-cm-thick Lucite in the x-ray beam path. This thickness of Lucite was used as it was found to be the median thickness range (4.5–4.99 cm) of the compressed breast from a random sample of 100 breast exams obtained from a population of 1400 patients. The antiscatter grid was not used while obtaining the images as it might provide a possible noise source, which might corrupt the measurement. In order to minimize scattered radiation affecting the measurement due to the removal of the antiscatter grid, the 4.5-cm-thick Lucite block was mounted on to the tube housing. In addition, the x-ray beam was collimated both at the tube port and at the surface of the detector using Pb (0.5 cm) so that only a 4 cm×4 cm area of the detector was irradiated. This enabled us to obtain our objective of achieving a realistic clinical spectrum without the measurement being affected by either excessive scattered radiation or the presence of structure from an antiscatter grid. The setup for NPS measurement is shown in Fig. 6. Fifteen dark image subtracted, flat field corrected, 256×256 ROIs were acquired as described previously. Before performing dark image subtraction and flat field correction, care was taken to avoid information loss due to truncation by converting the pixel intensity values to 32 bit floating point numbers from the original 16 bit digital values. A surface fit (like a ramp) to suppress background trends like heel effect was

performed on each ROI. The ensemble average of the squares of the magnitude of these 15 Fourier transformed 256×256 ROIs scaled as shown in Eq. (3) provided the 2D raw noise power spectrum, $NPS_{\text{raw}}(u, v)$.

The $NPS_{\text{raw}}(u, v)$ was obtained by

$$NPS_{\text{raw}}(u, v) = \frac{\langle |FT[\text{flat field}(x, y)]|^2 \rangle}{N_x N_y} \Delta_x \Delta_y, \quad (3)$$

where $\langle |FT[\text{flat field}(x, y)]|^2 \rangle$ represents the ensemble average of the squares of the magnitude of the Fourier transformed 256×256 ROIs, N_x and N_y are the number of elements in the x and y directions, respectively (which are equal and is 256 in this case), and Δ_x and Δ_y are the pixel pitch in x and y directions, respectively (which are equal and is $100 \mu\text{m}$ with this imager).

To compute noise equivalent quanta (NEQ) and DQE a 1D NPS curve was required. This was achieved by using the data in a thick slice comprised of eight lines on either side of both the u and v axes (excluding the axes). For each data value at (u, v) in this thick slice, the frequency value was computed as $\sqrt{u^2 + v^2}$ for the 1D NPS estimate. The final 1D NPS at each exposure level is the average of 8 (lines) $\times$ 2 (sides) $\times$ 256 data points (=4096 data values) grouped into frequency bins 0.04 mm^{-1} . The 1D $NPS_{\text{normalized}}(f)$ to be used for the DQE calculations was obtained by scaling the 1D $NPS_{\text{raw}}(f)$ for the mean signal by

$$NPS_{\text{normalized}}(f) = \frac{NPS_{\text{raw}}(f)}{(\text{mean signal of } 256 \times 256 \text{ ROI})^2}. \quad (4)$$

The mean signal of the 256×256 ROI is expressed in digital values.

The electronic noise present in the system was also estimated. The entire detector was covered with Pb (2 cm) and 15 images were acquired using the minimum possible exposure technique. The 2D $NPS_{\text{electronic}}(u, v)$ was estimated as per Eq. (3) at this minimum possible exposure technique with Pb, and the 1D $NPS_{\text{electronic}}(f)$ estimated by using a thick slice as described earlier. From this measurement, the noise contribution due to the x rays, $NPS_{x \text{ ray}}(f)$ was calculated at each exposure level as per Eq. (5), where $NPS_{\text{raw}}(f)$ is the raw NPS estimated as per Eq. (3) and $NPS_{\text{electronic}}(f)$ is the electronic noise of the system. The x-ray component of $NPS_{\text{raw}}(f)$ was computed as per Eq. (6)

$$NPS_{x \text{ ray}}(f) = NPS_{\text{raw}}(f) - NPS_{\text{electronic}}(f), \quad (5)$$

$$x - \text{ray component of } NPS_{\text{raw}}(f) = \frac{NPS_{x \text{ ray}}(f)}{NPS_{\text{raw}}(f)} \times 100\%. \quad (6)$$

In order to study the structured noise component or the presence of any varying nonstochastic noise, the 2D $NPS_{\text{subtracted}}(u, v)$ was estimated as per Eqs. (7) and (8).

Background suppression (ramp fit) was not performed for estimation of $NPS_{\text{subtracted}}(u, v)$. The 1D $NPS_{\text{subtracted}}(f)$ was obtained by using a thick slice of eight lines of data on either side of the axes as described earlier,

$$\text{residual}_i(x, y) = [\text{flood}_i(x, y) - \text{dark}_i(x, y)] - \frac{1}{n} \sum_{i=1}^n \text{flat field}_i(x, y), \quad (7)$$

$$NPS_{\text{subtracted}}(u, v) = \frac{\langle |\text{FT}(\text{residual}(x, y))|^2 \rangle}{(\text{mean signal of } 256 \times 256 \text{ ROI})^2 N_x N_y} \Delta_x \Delta_y. \quad (8)$$

C. NEQ and DQE measurement

The NEQ was computed as

$$NEQ(f) = \frac{MTF^2(f)}{NPS_{\text{normalized}}(f)}. \quad (9)$$

The NEQ of the system was computed for the four exposure levels. For the purpose of calculating the DQE of the digital imager, Eqs. (10) and (11) were used:

$$DQE(f) = \frac{MTF^2(f)}{NPS_{\text{normalized}}(f)q}. \quad (10)$$

and hence

$$DQE(f) = \frac{NEQ(f)}{q}. \quad (11)$$

where $MTF(f)$ is the modulation transfer function of the system; $NPS_{\text{normalized}}(f)$ is the normalized noise power spectrum of the imaging system; q is the number of x-ray photons incident on the detector per unit area; $NEQ(f)$ is the noise equivalent quanta of the imaging system and f is the spatial frequency. The only factor that needs to be determined is q .

Determination of q —Determination of q was done in three stages. First, the x-ray photon fluence per mR was curve fitted between the energy range of 5 to 35 keV from already published values⁴⁰ and is shown in Fig. 7. The photon fluence per mR, $Y(e)$, at energy (e) is best described by the polynomial:

$$Y(e) = 2.2128 + 33.514e + 89.23e^2 + 3.0588e^3 - 0.0239e^4 - 0.0006e^5 - 3 \times 10^{-7}e^6. \quad (12)$$

The x-ray spectral distribution, $q(e)$, was characterized by averaging 15 spectra obtained using a cadmium zinc telluride (CZT) based high resolution spectrometer (XR-100T-CZT, Amptek, Inc., USA). The x-ray spectrum was corrected for dead time losses and pile-up.⁴¹ Correction for the spectrometer energy response was not needed as the energy absorption efficiency of the 3-mm-thick CZT spectrometer is more than 99.9% for the energy range (5–35 keV) of the incident spectrum. The exposure (X) on the surface of the detector was measured under the same conditions as during the NPS measurement with a calibrated mammographic ionization chamber connected to MDH 1515 (RadCal Corp., USA) dosimeter. The precision at each exposure level was improved by averaging five measurements. The total number of photons incident per unit area of the detector at each exposure level was calculated as per Eq. (13). With the knowledge of q , the DQE(f) was calculated,

$$q = X \frac{\int q(e)Y(e)de}{\int q(e)de}. \quad (13)$$

III. RESULTS AND DISCUSSION

A. Presampling MTF

The measured presampling MTF is shown in Fig. 8. The presampling MTF measured both along the vertical and horizontal directions were identical. The presampling MTF was found to be 0.73, 0.42 and 0.28 at 2, 4, and 5 cycles/mm, respectively. Although the MTF of an imaging system is an important objective measure of the spatial resolution, this parameter alone may not be predictive of the overall performance of the system. Other metrics such as DQE as a function of the spatial frequency provide additional insight.

B. Noise power spectra

The linearity of the system was measured and is shown in Fig. 9. From the linearity measurements the sensitivity of the system was found to be 16.324 digital values/mR/pixel. The 2D NPS obtained at 1.3, 7.1, 14.5, and 32.8 mR are shown in Figs. 10(a), 10(b), 10(c), and 10(d), respectively. The noise power at the intersection of the u and v axes are much higher in magnitude and hence this point has been blanked for display purposes. The images are displayed in a black and white scheme where the transition point is set at the midpoint of the minimum and maximum of the 2D NPS images. The 2D NPS does not show the presence of any off-axis noise peaks. The 1D NPS_{raw} at four exposure levels of 1.3, 7.1, 14.5, and 32.8 mR are shown in Fig. 11. The electronic noise present in the system is also shown in Fig. 11. The 1D NPS_{raw} demonstrates an increase in noise with increasing exposure as the photon noise increases with increasing exposure. The integral of the NPS at each exposure was confirmed to be identical to the rms variance of the 256×256 ROI. Figure 12 shows the x-ray component of the total NPS calculated as per Eq. (6) at the four exposure levels. Even at a low exposure of 1.3 mR, the x-ray component was dominant (greater than 60% of the total NPS at 5 cycles/mm and approximately 80% of the total NPS at ~0 cycle/mm). Figure 13 suggests that there is no appreciable structure noise or varying non-stochastic noise at exposures of 1.3 and 32.8 mR as the NPS_{subtracted} and NPS_{normalized} are identical.

C. NEQ and DQE

The NEQ of the system at four exposure levels are shown in Fig. 14. The Mo–Mo spectrum incident on the detector transmitted through 4.5 cm of Lucite and the breast support plate recorded with a high resolution spectrometer is shown in Fig. 15. From this spectral distribution and Fig. 7, the photon flux incident on the detector was determined to be 0.533×10^5 photons/mm²/mR. The DQE of the system at four exposure levels is shown in Fig. 16. To demonstrate the exposure dependence of the DQE of the system, DQE (0.2 cycle/mm), DQE(1 cycle/mm), DQE(2 cycles/mm), DQE(3 cycles/mm), and DQE(5 cycles/mm) are plotted as a function of the incident exposure in Fig. 17. The plot indicates that the DQE of the system increases with increasing exposure, and reaches a constant value at about 15 mR. The lower values of DQE at low exposures are primarily due to the contribution of electronic noise in the system. The DQE (~0 cycle/mm) was found to be 0.4, 0.48, 0.54, and 0.55 at incident exposures of 1.3, 7.1, 14.5, and 32.8 mR, respectively.

D. Discussion

Metrics such as MTF and DQE have been widely used to describe the performance characteristics of imaging systems. A comparison of the flat-panel imager with other imaging systems such as screen-film systems to show the general trends could provide additional information as to the advantages and limitations of the flat-panel imager. Nishikawa and Yaffe⁴² have evaluated various mammographic screen-film systems in the past. More recently, Bunch⁴³ has also evaluated the MTF and DQE of two widely used mammographic screen-film systems. Their results show a maximum DQB(~0) of 0.35 compared with 0.55 measured with the flat panel imager. The improved DQE of the flat panel imager at low and midfrequencies can be particularly advantageous in the imaging of low-contrast soft tissue lesions.^{44,45} Their results also indicate that the spatial resolution is much higher with screen-film systems^{42,43} compared to the flat-panel imager, but an increased film noise at high frequencies have also been observed.

Previous laboratory studies²³ in digital mammographic imaging using different technology have suggested that even with lower spatial resolution, lesion detectability, including microcalcifications can be improved by contrast enhancement of digital data. Prior work with this flat-panel imager has demonstrated a high dynamic range.⁴⁶ Clinical images with the current prototype demonstrate encouraging results for visualization of soft tissue anatomy and calcifications.^{47–49} The clinical efficacy in terms of sensitivity and specificity is the subject of a different investigation, which is currently in progress.⁴⁷

IV. CONCLUSIONS

A consistent set of image quality measurements was performed characterizing the full field amorphous silicon-based flat panel imager for mammographic applications. The flat panel imager did not exhibit any appreciable structured noise or varying nonstochastic noise component at the tested exposure levels. The response of the imager was linear and exhibited high sensitivity under tested exposure conditions. The flat panel imager demonstrated good dose efficiency within the tested exposure range.

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Fig. 1.
Schematic of the amorphous silicon detector array.

Fig. 2. Experimental setup for MTF measurement. The area surrounding the 10 μm slit was covered with Pb (0.5 cm thick).

Fig. 3.
Illustration of slit image correction for variations in slit width.

Fig. 4.
The pixel amplitudes along the anode–cathode axis used for determining the number of rows of data needed to obtain a finely sampled LSF.

Fig. 5.
Finely sampled LSF.

Fig. 6.

Experimental setup for NPS measurement where a 4 cm×4 cm area of the detector centered at 4 cm from the chest wall edge was irradiated. Lead collimation at the tube port and at the detector surface reduced excessive scatter.

Fig. 7.

Curve fitted x-ray photon fluence per mR between the energy range of 5 and 35 keV obtained from published values.

Fig. 8.
The presampling MTF of the full field flat panel α :Si imager.

Fig. 9.
Linearity of the system. The data points represent the mean intensity and the error bars represent the standard deviation from this mean value.

Fig. 10.

The 2D NPS obtained at 1.3, 7.1, 14.5, and 32.8 mR are shown in (a), (b), (c), and (d), respectively. The intersection of the axes has been masked for display purposes. The images are displayed in a black and white scheme, with the transition point set at the mean of the ROI.

Fig. 11.

The 1D noise power spectra (NPS_{raw}) at four exposure levels of 1.3, 7.1, 14.5, and 32.8 mR are shown. The electronic noise is also shown.

Fig. 12.
The x-ray component of NPS_{raw} at four exposure levels of 1.3, 7.1, 14.5, and 32.8 mR are shown.

Fig. 13.
The 1D $NPS_{\text{normalized}}$ and $NPS_{\text{subtracted}}$ obtained at 1.3 and 32.8 mR.

Fig. 14.
The NEQ of the system at four exposure levels.

Fig. 15.
The Mo–Mo spectra incident on the detector transmitted through 4.5 cm of Lucite and the breast support plate, recorded with a high resolution spectrometer for calculation of q .

Fig. 16.

The DQE of the system at four exposure levels. Data points are curve fitted with a sixth-order polynomial for clarity. To demonstrate the goodness of fit, data points at an exposure of 1.3 mR are shown.

Fig. 17.
DQE of the system plotted as a function of incident exposure.

Table I

Amorphous silicon-based flat panel detector specifications.

Flat panel image area	18 cm×23 cm
Pixel matrix	1800×2304
Pixel size	100 μ m
Scintillator	CsI:Tl