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Mammographic imaging with a small format CCD-based digital cassette: Physical characteristics of a clinical systema

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Abstract

The physical characteristics of a clinical charge coupled device (CCD)-based imager (Senovision, GE Medical Systems, Milwaukee, WI) for small-field digital mammography have been investigated. The imager employs a MinR 2000^{TM} (Eastman Kodak Company, Rochester, NY) scintillator coupled by a 1:1 optical fiber to a front-illuminated 61×61 mm CCD operating at a pixel pitch of 30 microns. Objective criteria such as modulation transfer function (MTF), noise power spectrum (NPS), detective quantum efficiency (DQE), and noise equivalent quanta (NEQ) were employed for this evaluation. The results demonstrated a limiting spatial resolution (10% MTF) of 10 cy/mm. The measured DQE of the current prototype utilizing a 28 kVp, Mo–Mo

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spectrum beam hardened with 4.5 cm Lucite is ~40% at close to zero spatial frequency at an exposure of 8.2 mR, and decreases to ~28% at a low exposure of 1.1 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)

I. INTRODUCTION

In recent years, advances in screen-film mammography and film processing techniques have contributed to significant improvements in mammographic image quality. While screen-film techniques provide a powerful tool for initial detection and subsequent follow-up of a suspicious area, they present significant limitations in detecting very subtle soft tissue lesions, especially in the presence of dense glandular tissue. Some of the fundamental limitations of screen-film mammography, particularly with respect to contrast and noise, have been discussed in several studies.^{1,2} Consequently, attempts have been made to explore the potential of electronic detection as an alternate detection technique. Systems based on electronic detection have the theoretical capability of overcoming certain fundamental limitations of screen-film systems. The potential advantages of electronic detection include high detection efficiency, high dynamic range, capability for contrast enhancement,³ and post processing capabilities including computer-aided diagnosis.^{4–9} Further, direct electronic acquisition enables the exploration of novel imaging techniques such as tomosynthesis,^{10,11} dual-energy mammography,^{12,13} and digital subtraction imaging.¹⁴ In the past, investigators have used different modes of electronic detection technology to gain insight into electronic mammography, commonly referred to as digital mammography.¹⁵ Early evaluations have used image intensifiers and subsequently slot-scanned systems^{1,2,16,17} with charge-coupled devices (CCDs) and CCDs with fiberoptic tapers.¹⁸ The potential for utilizing CCD-based imagers for small-field digital mammography was described by Karellas *et al.*¹⁸ and, now, the use of CCDs for core biopsies has become common practice. The use of core biopsies has been increasing in the past 10 years and a number of open surgical excisions are being replaced by these minimally invasive procedures. Although screen-film systems produce excellent image quality for these procedures, the film development process severely hinders fast display of acquired images, resulting in patient discomfort. The recent adaptation of CCD technology has enabled electronic acquisition of mammographic images during these procedures quickly and efficiently. Core biopsy procedures performed with an electronic imaging device can reduce the duration of the procedure and patient discomfort. The study by Dershaw *et al.*¹⁹ demonstrated the reduced duration for completion of needle localization studies when using digital technology. Moreover, digital imaging systems for mammographically guided digital stereotactic breast biopsy have an important advantage over screen-film systems in that they provide a digital output that can be used for quantitative analysis.²⁰ Observer performance

comparison of digital radiograph systems for stereotactic breast needle biopsy has also been reported in the past.²¹

The first generation of these devices employed either a lens or a fiberoptic taper to couple the scintillator with the CCD. Field coverage of 5×5 cm to 6×6 cm is typical. Although this is a very restricted field of view, it is considered adequate for most localization and core biopsy procedures. The spatial resolution of these first generation devices was lower than that of the screen-film systems, and the limited optical coupling efficiency due to demagnification between the x-ray scintillator and CCD presented a significant challenge in attaining high detective quantum efficiency (DQE). The geometric demagnification between the scintillator and CCD reduces spatial resolution by virtue of the geometry of the optics. The light loss due to the demagnification reduces the optical signal to the CCD, and therefore contrast and dose efficiency are negatively affected. With the present day ability to manufacture large CCDs (6×6 cm, typically), the fiberoptic tapers or lens coupling which pose serious limitations can be overcome with a straight optical fiber. While this approach of coupling the CCD with the scintillator using a straight optical fiber provides the theoretical capability of improved optical efficiency, detailed experimental characterization of the physical properties of such imagers under realistic mammographic conditions have not been reported in the past. This study characterizes the image quality parameters of a CCD-based clinical imaging system (Senovision, GE Medical Systems, Milwaukee, WI) which employs a MinR 2000™ (Eastman Kodak Company, Rochester, NY) scintillator coupled by a 1:1 optical fiber to a front-illuminated 61×61 mm CCD.²² See Table I.

II. METHODS AND MATERIALS

The CCD used in this imager is a full-frame area image sensor with a matrix array of 4096 horizontal by 4097 vertical detector elements (pixels). The pixel pitch and spacing is 15 μm . The imaging array is operated in the multipinned phase (MPP)²³ mode. In this mode, the dark current is decreased down to 25 pA/cm² at room temperature of 25 °C. The dark current is further minimized to ~10 pA/cm² by cooling the CCD to the operating temperature of 12 °C by a liquid circulation system. The CCD was manufactured using 2.5 micron design rules. The single-metal, triple-poly process allows a layout with small pixel geometries and few array blemishes. Incident photons pass through a transparent polycrystalline silicon gate structure, creating electron hole pairs. The resulting photoelectrons are collected in the pixels during the integration period. The amount of charge accumulated in each pixel is a linear function of the localized incident illumination density and integration period. The pixel structure is made up of contiguous CCD elements with no voids or inactive areas. In addition to sensing light photons, these elements are used to shift image data vertically. Consequently, x rays must not be detected during this transfer period. The full-frame architecture of the CCD provides image data as a sequential readout of 4097 lines, each containing 4096 pixels. At the end of the integration period, a three-phase clocking mechanism is utilized to transfer charge vertically through the CCD array to the horizontal readout register. A channel stop region between vertical columns separates the columns to prevent charge migration. The imaging area is divided into four quadrants and each quadrant may be clocked independently, if desired. The CCD may be clocked such that the full array is read out from any one of the four output amplifiers. The present readout

mode utilizes only one amplifier and reads out the full array through this amplifier. The last clocked gate in the horizontal registers is larger than the other gates to facilitate binning (grouping of adjacent pixels prior to readout) the charge packets horizontally. The CCD has four, dual field-effect transistor (FET), floating diffusion output amplifiers, with a reset metal-oxide-semiconductor field-effect transistor (MOSFET) tied to the input gate. Charge packets are clocked to a precharged capacitor whose potential changes linearly in response to the number of electrons delivered. This potential is applied to the input gate of the amplifier producing a signal at the output. The capacitor is reset to a precharge level using the reset MOSFET, prior to the arrival of the next charge packet, except when horizontally binning. The output from the CCD is connected to an external load resistor to ground. The CCD array is operated in a 2×2 pixel binned mode to provide a full-frame image area of 2048×2048 pixels with a pixel pitch of $30 \mu\text{m}$. Vertical binning is achieved by transferring two lines of charges from 4096 pixels onto the horizontal register. Horizontal binning is achieved by transferring two charge packets onto the last clocked, larger gate of the horizontal register and resetting the capacitor to the precharge level after the arrival of two charge packets. The charge packets' readout through the output amplifiers are digitized to 12 bits, providing digital values in the range of 0 to 4095. The schematic of the detector is shown in Fig. 1. The detector is also designed to fit the 18×24 cm cassette tray of mammographic systems providing an easy transition from a screen-film system to a digital system. The imager is integrated with a high-frequency x-ray generator (Senographe DMR, GE Medical Systems, Milwaukee, WI). This system uses a selectable dual track target, either molybdenum or rhodium, with selectable filtration of molybdenum or rhodium. All measurements were performed at 28 kVp with a Mo/Mo target/filter combination.

A. Linearity

In order to study the signal and noise performance through Fourier components, the detector has to be linear and stationary.^{24–26} The linear response of the system was measured by averaging the pixel intensity over the entire 2048×2048 image at various exposure levels. The images were obtained without the antiscatter grid in place, as the antiscatter grid is not used during clinical stereotactic localization studies. In order to measure the linearity with a clinically relevant spectrum, a 4.5 cm thick Lucite block was used in the x-ray beam path. The Lucite block was mounted onto the tube housing to reduce scattered radiation. The experimental setup for measuring the linear response is shown in Fig. 2. The sensitivity (signal per pixel/mR) of the system was calculated to be the slope of the linear response curve.

Under the condition of nominally uniform exposure to the detector, the stationary property can be reasonably assumed. Also, the assumption of ergodicity (which implies stationarity)^{25,26} has been made to facilitate ensemble averaging of noise data.

B. Presampling MTF measurement

The presampling modulation transfer function (MTF) was measured based on the slanted-slit technique described by Fujita *et al.*²⁷ The experimental procedure for measuring the same has also been described in detail by Dobbins *et al.*²⁸ Dobbins²⁹ has also described the effect of undersampling in detail. The specific methodology employed for measurement of

the presampling MTF is identical to that used with the amorphous silicon-based imager, which was presented previously.³⁰ Hence, only specific attributes to the measurement procedure employed with this system alone would be addressed in this paper. The experimental setup is shown in Fig. 3. The presampling MTF was measured both along the anode–cathode axis and perpendicular to the anode–cathode axis. As an example, the methodology used for measuring the presampling MTF perpendicular to the anode–cathode axis is presented alone. A dark-subtracted image of a 10 micron slit oriented at a slight angle to the anode–cathode axis was acquired. Since imperfections along the edges of the slit result in variations in slit width, a normalization scheme was used. In this scheme, the amplitude of each pixel (x) was divided by the sum of the amplitudes of all pixels in a line that is oriented perpendicular to the anode–cathode axis that includes the pixel “ x .” This normalization scheme is feasible only if the assumptions that the slit width is approximately constant over the region used for obtaining the line spread function (LSF) and that the signal spreading is approximately equal along each line are made. These assumptions were verified by measuring the presampling MTF from several locations of the slit, the presampling MTF varied by less than 0.5%. In order to synthesize a finely sampled LSF, the adequate number of individual LSFs needs to be determined. This was achieved by plotting adjacent lines of pixel amplitudes along the anode–cathode axis as shown in Fig. 4. The separation between the two points of intersection determines the adequate number of LSFs required for obtaining a finely sampled LSF. The composite LSF was synthesized by using 33 individual LSFs. The composite LSF was normalized to a peak value of 1 and is shown in Fig. 5. The Fourier transform (FT) of the composite 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.

C. NPS measurement

The difficulties in measuring the noise power spectrum (NPS) of digital systems^{24–26,28–32} have been described. The NPS can be calculated via the auto correlation function (indirect method) or by the Fourier transform of the image (direct method). With the advent of the fast Fourier transform and fast computers, the indirect method has largely been replaced with the direct method.²⁶ The NPS measurements reported in this paper were performed with the direct method. The typical assumption of ergodicity, usually made with radiographic images, has been made to facilitate this analysis. The presampling NPS cannot be directly measured using fine sampling techniques such as those employed to measure the presampling MTF, because the phases of the Fourier components of the image noise are random²⁷ and hence the measured NPS is inherently aliased. The experimental methodology used for NPS measurement is similar to that presented earlier.³⁰ The noise power spectra were determined at five exposure levels and were obtained with 4.5 cm thick Lucite block mounted on to the tube housing and without the anti-scatter grid. In order to minimize scattered radiation, the x-ray beam was restricted both at the tube port and at the detector surface using Pb (0.5 cm), so that only a central 4×4 cm area of the detector is 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 anti-scatter grid. The setup for NPS measurement is shown in Fig. 6.

Ten dark image subtracted, flat-field corrected images of 2048×2048 pixels were acquired at each exposure level. The central 512×512 pixel matrix was obtained from each image. The 512×512 pixel matrix obtained was subdivided into four 256×256 ROIs for estimation of the noise power spectrum. Hence, a total of 40 (=4×10 images) regions of interest (ROIs) at each exposure level was used to determine the NPS. Problems associated with background trends such as from the heel effect can corrupt the noise spectrum and provide artificially inflated values,^{28,32} along the axes. Hence, we surface (ramp) fitted each ROI and subtracted these background trends. The 2D Fourier transform of each of the 40 ROIs was performed. The ensemble average of the squares of the magnitude of these 40 Fourier transformed ROIs were scaled as shown in Eq. (1) to obtain the two-dimensional (2D) raw noise power spectrum, $NPS_{raw}(u, \nu)$.

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

$$NPS_{raw}(u, \nu) = \frac{\langle |FT(u, \nu)|^2 \rangle}{N_x \cdot N_y} \cdot \Delta_x \cdot \Delta_y, \quad (1)$$

where $\langle |FT(u, \nu)|^2 \rangle$ is 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 are 256 in this case), and Δ_x and Δ_y are the pixel pitch in the x and y directions, respectively (which are equal and are 30 μm in this imager).

Although the ramp fit was successful in suppressing these background trends, it did not completely eliminate them. Hence, the data values directly on the axes were avoided while estimating the 1D NPS from the 2D NPS.

For the exposure levels, which demonstrated a nominal radial symmetry, the assumption of radial symmetry has been made. The 1D NPS curve required for estimation of NEQ and DQE was obtained by radially averaging the data in a thin slice comprised of eight lines on either side of both the u and ν axes (excluding the axes). For each data value at (u, ν) in this slice, the frequency value (f) was computed as $\sqrt{u^2 + \nu^2}$ for the 1D NPS estimate.²⁸ The final 1D NPS at each exposure level is the average of 8(lines)×2(axes)×256 data points (=4096 data values) grouped into frequency bins of 0.13 mm^{-1} .

For the exposure levels at which the 2D NPS did not demonstrate radial symmetry, the NPS along the u and ν axes were extracted separately. For these exposure levels, although the 2D NPS does not demonstrate radial symmetry, a nominal radial symmetry has been assumed within the thin slice used along either of the axes, to facilitate radial averaging of the data within this thin slice, with the frequency value computed as $\sqrt{u^2 + \nu^2}$. For these exposure levels, the 1D NPS at each exposure level is obtained as the average of 8(lines)×1(axis)×256 data points (×2048 data values) grouped into frequency bins of 0.13 mm^{-1} .

The 1D $NPS_{normalized}(f)$ used for the estimating the DQE was obtained by dividing the 1D $NPS_{raw}(f)$ by the mean signal as shown in Eq. (2).

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

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

D. NEQ and DQE measurement

The NEQ was computed as

$$NEQ(f) = \frac{MTF^2(f)}{NPS_{\text{normalized}}(f)}. \quad (3)$$

The NEQ of the system was computed for the five exposure levels. For the purpose of calculating the DQE of the digital imager, the equation below was used:

$$DQE(f) = \frac{MTF^2(f)}{NPS_{\text{normalized}}(f) \cdot q}, \quad (4)$$

and, hence

$$DQE(f) = \frac{NEQ(f)}{q}, \quad (5)$$

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 per unit area incident on the detector, $NEQ(f)$ is the noise equivalent quanta of the imaging system, and f is the spatial frequency.

For the exposure levels that did not demonstrate radial symmetry, there were two 1D noise spectra, one along each of the two axes. The noise spectra used to represent the noise power at a particular exposure level for calculation of the $DQE(f)$ was selected based on (i) the area under the noise spectra closest to the measured rms variance of the ROI, and (ii) the noise spectra that demonstrates the falloff trend of $NPS(f) \propto MTF^2(f)$. As noted by Lubberts,³³ the $NPS(f)$ does not follow this trend at higher spatial frequencies. Hence, the selection between NPS obtained along the u and v axes was based on the falloff trend up to the midfrequency of 6 cycles/mm.

1. Determination of q —Determination of q was performed using the recorded x-ray spectral shape, curve fit of the published values of photons incident per mR at each energy bin,³⁴ and the measured exposure onto the detector.³⁰ The incident x-ray spectra were recorded using a collimated, high-resolution, cadmium zinc telluride (CZT) based spectrometer.³⁵ The Mo–Mo spectrum was obtained by averaging 15 acquisitions of 100 mAs each. The spectrum was corrected for dead time losses and peak pileup.³⁶ With the knowledge of q , the spatial frequency dependent $DQE(f)$ was estimated.

III. RESULTS AND DISCUSSION

The measured linear response of the system is shown in Fig. 7. The error bars represent ± 1 standard deviation from the mean of the 2048 $\times$ 2048 image. From the slope of this linear response curve, the sensitivity was determined to be 19.06 digital values per pixel/mR.

A. Modulation transfer function

The measured presampling MTF is shown in Fig. 8. The presampling MTF measured along the anode–cathode axis and perpendicular to the anode–cathode axis were identical. Although the MTF of an imaging system is an important objective measure of the spatial resolution and the signal transfer characteristics, 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 2D NPS obtained at 1.1, 8.2, 14.2, 32.2, and 39.9 mR are shown in Figs. 9(a), 9(b), 9(c), 9(d), and 9(e), respectively. Each 2D NPS has been individually adjusted to provide maximum perceptibility. The 2D NPS are displayed in a black–white scheme with the transition point set at the mean intensity value of each 2D NPS. The noise power attributable to the off-axis noise peaks is small relative to overall noise power, and is increasingly true at higher exposures as the contribution of the x-ray quantum noise increases. While the 2D NPS at low to mid exposures demonstrate reasonable radial symmetry, at high exposures the 2D NPS demonstrate increasingly elliptical shape with increased exposures. The noise power at high exposures along the v axis is significantly higher than the noise power along the u axis, as shown in Figure 10 for an example exposure level of 39.9 mR. The increased noise power is along the vertical direction (parallel shift) of the CCD and may be associated with the charge transport properties of the CCD at high signal amplitudes. The normalized 1D noise power spectra ($\text{NPS}_{\text{normalized}}$) obtained from a thin slice of the 2D NPS at five exposure levels of 1.1, 8.2, 14.2, 32.2, and 39.9 mR are shown in Fig. 11. For the exposure levels of 1.1, 8.2, and 14.2 mR, the 1D NPS represent the average of thin slices located immediately adjacent and parallel to both the u and v axes. The integral of the NPS at each of these three exposures was confirmed to be identical to the rms variance of the 256 $\times$ 256 ROI. The 1D NPS also demonstrated an $\text{MTF}^2(f)$ falloff trend up to 6 cycles/mm.

For the exposure levels of 32.2 and 39.9 mR, the 1D NPS represent the average of thin slices located immediately adjacent and parallel to the u axis only, due to lack of radial symmetry. The 1D NPS along the u axis was selected to represent the noise spectra at 32.2 and 39.9 mR, as the integral of the NPS was within 2% of the rms variance of the 256 $\times$ 256 ROI and demonstrated an $\text{MTF}^2(f)$ falloff trend up to 6 cycles/mm.

C. NEQ and DQE

The NEQs of the system at five exposure levels are shown in Fig. 12. The Mo–Mo spectrum incident on the detector after transmitting through 4.5 cm of Lucite is shown in Fig. 13. From the spectral shape, the photon fluence incident on the detector was determined to be 5.34×10^4 photons/mm²/mR. The DQE of the system at five exposure levels is shown in Fig.

14. For the exposure levels of 32.2 and 39.9 mR, which did not demonstrate a radially symmetric 2D NPS, the $DQE(f)$ was computed with the $NPS(f)$ extracted along the u axis, while at the other exposure levels the $DQE(f)$ was computed with the $NPS(f)$ extracted along both the u - and v axes. To demonstrate the exposure dependence of the DQE of the system, $DQE(0.2 \text{ cy/mm})$, $DQE(2 \text{ cy/mm})$, $DQE(5 \text{ cy/mm})$, $DQE(10 \text{ cy/mm})$, and $DQE(15 \text{ cy/mm})$ are plotted as a function of the incident exposure in Fig. 15. The slight decrease in DQE at high exposures can probably be associated with the granular noise³⁷ due to the scintillator.

IV. CONCLUSION

A consistent set of image quality measurements was performed characterizing the small-field CCD-based digital cassette for mammographic applications. The $DQE(\sim 0)$ was measured to be 0.29, 0.40, 0.40, 0.37, and 0.36 at incident exposures of 1.1, 8.2, 14.2, 32.2, and 39.9 mR, respectively. The presampling MTF was found to be 0.78, 0.40, and 0.10 at 2, 5, and 10 cy/mm, respectively. The DQE of the CCD imager was found to be comparable to existing screen-film systems in the frequency range of zero to 10 cy/mm.³⁸

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Fig. 1.
Schematic of the CCD-based mammographic detector.

Fig. 2. Experimental setup for linearity measurement. The 4.5 cm thick Lucite block was mounted on to the tube housing to reduce excessive scatter.

Fig. 3.
Experimental setup for MTF measurement. The area surrounding the 10 micron slit was covered with Pb (0.5 cm thick).

Fig. 4.

The pixel amplitudes along the anode–cathode axis used for determining the adequate number of individual LSFs required for synthesizing a finely sampled LSF. Based on the separation between the two points of intersection, 33 individual LSFs were required for obtaining the finely sampled LSF.

Fig. 5.
Finely sampled composite LSF. The spacing between adjacent points is 0.91 microns.

Fig. 6. Experimental setup for NPS measurement, where a central 4×4 cm area of the detector was irradiated. Lead collimation at the tube port and at the detector surface reduced excessive scatter.

Fig. 7. Linearity of the system. The data points represent the mean intensity of 2048×2048 image. The error bars represent ± 1 standard deviation from this mean value.

Fig. 8.
The presampling MTF(f) of the small-field CCD-based digital cassette.

Fig. 9.
The 2D NPS obtained at 1.1, 8.2, 14.2, 32.2, and 39.9 mR are shown in (a), (b), (c), (d), and (e) respectively.

Fig. 10.

As an example, the 1D $NPS_{\text{normalized}}$ at a high exposure of 39.9 mR obtained along the u axis and v axis are shown. The noise power along the v axis is significantly higher than the noise power along the u axis.

Fig. 11.

The 1D normalized noise power spectra ($NPS_{\text{normalized}}$) at five exposure levels are shown. The 1D $NPS_{\text{normalized}}$ at 1.1, 8.2, and 14.2 mR were extracted from thin slices immediately adjacent and parallel to both the u and v axes. The 1D $NPS_{\text{normalized}}$ at 32.2 and 39.9 mR were extracted from thin slices immediately adjacent and parallel to the u axis only.

Fig. 12.

The NEQ of the system at five exposure levels. NEQ(f) at 1.1, 8.2, and 14.2 mR were obtained from the NPS(f) extracted along both the u and v axes and the NEQ(f) at 32.2 and 39.9 mR were obtained from the NPS(f) extracted along the u axis only.

Fig. 13.
The Mo–Mo spectrum incident on the detector after transmission through 4.5 cm of Lucite, recorded with a cadmium zinc telluride-based high-resolution spectrometer, for calculation of q .

Fig. 14.

The DQE of the system at five exposure levels. Data points are curve fitted with a fourth-order polynomial for clarity. To demonstrate the goodness of fit, data points at an exposure of 1.1 mR are shown. DQE(f) at 1.1, 8.2, and 14.2 mR were obtained from the NPS(f) extracted along both the u and v axes, and the DQE(f) at 32.2 and 39.9 mR were obtained from the NPS(f) extracted along the u axis only.

Fig. 15.
DQE of the system plotted as a function of incident exposure.

Table I

CCD-based mammographic detector specifications.

CCD image area	61×61 mm
Pixel matrix	4096×4097
Pixel size	15 μm
Scintillator	MinR 2000 ^{TMa}
Operating temperature	12 °C
Pixel binning	2×2
Binned pixel size	30 μm
Binned pixel matrix	2048×2048

^a_{TM}Eastman Kodak Company, Rochester, NY.