Modeling, validation and application of a mathematical tissue-equivalent breast phantom for linear slot-scanning digital mammography

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Abstract

This paper presents a mathematical tissue-equivalent breast phantom for linear slot-scanning digital mammography. A recently developed prototype linear slot-scanning digital mammography system was used for model validation; image quality metrics such as image contrast and contrast-to-noise ratio were calculated. The results were in good agreement with values measured using a physical breast-equivalent phantom designed for mammography. The estimated pixel intensity of the mathematical phantom, the analogue-to-digital conversion gain and the detector additive noise showed good agreement with measured values with correlation of nearly 1. An application of the model, to examine the feasibility of using a monochromatic filter for dose reduction and improvement of image quality in slot-scanning digital mammography, is presented.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

Quantitative image quality parameters such as modulation transfer function (MTF), noise power spectrum (NPS), detective quantum efficiency (DQE), signal-to-noise ratio (SNR) and figure of merit (FoM) are well known in the assessment of digital mammography detector performance (Siewerdsen and Antonuk 1998, Siewerdsen et al 1998, Stierstorfer and Spahn 1999, Siewerdsen and Jaffray 2000, Tkaczyk et al 2001, Cunningham 1994, Cunningham et al 2002, Neitzel et al 2001). Several test phantoms have been developed for performance assessment in digital mammography, such as the ACR (Barnes and Hendrick 1994), TRO (MAM) (Cowen and Coleman 1990) and IQI (Hessler et al 1985) test phantoms. Images produced by these test phantoms are usually subjectively assessed by a human observer.
The differences among human observers in the visibility thresholds of structures such as microcalcifications result in variations in perceived system performance. This limitation gives rise to a need for objective assessment. Several objective methods that have been developed to quantify system performance use computational software to emulate human detection, and are comparable with but more consistent than human observers (Chakraborty and Eckert 1995, Smith et al 1998, Verdun et al 1996, Veldkamp and Karssemeijer 1994, Pachoud et al 2004). These objective methods mainly use a physical phantom and depend on the sensitivity of the phantom to x-ray exposure, which will affect the quantitative assessment of image quality. This approach requires acquisition of a large number of phantom images, to quantify image quality. These limitations indicate a need for a mathematical phantom with which to simulate phantom images for system design and optimization. Several 2D and 3D mathematical simulations of breast tissues and microcalcifications have been developed. Bakic et al (2002) presented a model to simulate 2D mammograms based on simulated 3D anatomy. Hunt et al (2005) and Dance et al (2005) used a voxelized phantom to estimate mean glandular dose for breast dosimetry. Reiser et al (2006) developed an analytical phantom for tomosynthesis. Bliznakova et al (2003) simulated a 3D non-compressed breast containing tissues of different size, shape and composition, in order to generate images resembling real mammograms. They assumed the only source of noise to be the natural fluctuation of the photon flux (quantum mottle).

This paper presents a tissue-equivalent mathematical breast phantom, its validation for slot-scanning digital mammography and an examination of its use to determine the feasibility of using a monochromatic filter in digital mammography. Some structures in the mathematical phantom were simulated similarly to those in the physical tissue-equivalent phantom designed for mammography (Model 011 A) manufactured by CIRS (Computerized Imaging Reference Systems, Norfolk, VA, USA). The model was validated by comparing images acquired of the physical breast phantom, using a prototype linear slot-scanning digital mammography system, with simulated phantom images. The mathematical phantom was used to examine the feasibility of using a monochromatic filter for dose reduction and improvement of image quality in slot-scanning digital mammography. All programs were coded using Matlab (Matlab\textsuperscript{TM}, The MathWorks, Natick, MA).

2. Methods

The linear slot scanning mammography x-ray prototype system used for model validation uses the same linear slot scanning technology as the Lodox Statscan low-dose digital x-ray machine (Beningfield et al 2003). The system consists of an x-ray source which is mounted on a C-arm, utilizing a narrow fan beam that scans the patient by a horizontal movement along the Z-direction in synchronism with a narrow slotted Hamamatsu detector.

A successful x-ray phantom should mimic the attenuation properties of the tissues it represents. The mathematical breast phantom was simulated using breast tissue equivalent material of 50:50 glandular/adipose tissue. As shown in figure 1 and table 1, five areas containing structures simulating the absorption of 100% glandular, 70:30 glandular/adipose, 50:50 glandular/adipose, 30:70 glandular/adipose and 100% adipose tissues, four hemispheric regions and specks simulating tumor masses and microcalcifications of different sizes were present in the phantom. The sizes were selected to include the minimum and maximum tumor masses and microcalcifications used in the physical tissue-equivalent phantom designed for mammography (Model 011 A) manufactured by CIRS (Computerized Imaging Reference Systems, Norfolk, VA, USA). The mathematical phantom image was modeled using the following steps: modeling the pixel intensity; estimating the object contrast; modeling the detector noise and modeling the total system resolution.
Figure 1. Areas within the phantom simulating (1) 100% glandular, 70:30 glandular/adipose, 50:50 glandular/adipose, 30:70 glandular/adipose and 100% adipose tissues, from left to right; (2) four hemispheric regions; (3) microcalcifications of different sizes (CaCO$_3$); (4) line pair tools for measuring the system resolution and (5) nylon fibers of different sizes.

Table 1. Mathematical phantom specifications.

<table>
<thead>
<tr>
<th>Structures</th>
<th>Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td>(1) Step wedge</td>
<td>100% glandular</td>
</tr>
<tr>
<td></td>
<td>70% glandular</td>
</tr>
<tr>
<td></td>
<td>50% glandular</td>
</tr>
<tr>
<td></td>
<td>30% glandular</td>
</tr>
<tr>
<td></td>
<td>100% adipose</td>
</tr>
<tr>
<td>(2) Hemispheric masses</td>
<td>75% glandular/25% adipose, thickness</td>
</tr>
<tr>
<td></td>
<td>1 mm</td>
</tr>
<tr>
<td></td>
<td>2 mm</td>
</tr>
<tr>
<td></td>
<td>3 mm</td>
</tr>
<tr>
<td></td>
<td>4 mm</td>
</tr>
<tr>
<td>(3) Calcification (CaCO$_3$) grain size</td>
<td>0.122, 0.170 and 0.4 mm</td>
</tr>
<tr>
<td>(4) Line pair tools</td>
<td>7 lp mm$^{-1}$</td>
</tr>
<tr>
<td>(5) Nylon fibers</td>
<td>1.2, 0.8, 0.6, 0.4 and 0.3 mm</td>
</tr>
</tbody>
</table>

2.1. Modeling the pixel intensity in digital units

The pixel intensity ($S_{Te}$) of the mathematical breast phantom in digital units (DU) is proportional to the photoelectron intensity ($S_e$) generated from the x-ray photons incident on the detector and the added signal (dark signal) due to thermal electrons generated in the charge-coupled device (CCD) ($D_s$). The pixel intensity is corrected by subtracting the dark signal from the total signal. The corrected signal is then proportional to the photoelectron intensity:

$$S_{Te} - D_s \propto S_e.$$  \hspace{1cm} (1)

The signal (photoelectrons, $S_e$) generated by the x-ray detector from the attenuated x-ray photons depends on the following:

(i) The efficiency of the detector scintillator, characterized by its quantum efficiency (QE$_{Si}$).
(ii) The conversion of x-ray energy into light photons, characterized by the scintillator gain (DetGain).
(iii) The efficiency of transfer of quanta to the CCD, characterized by the detector coupling efficiency (CE).
(iv) The integration of quanta by the CCD, characterized by the active pixel area of the CCD ($A_P$).

Using linear cascaded system analysis and the propagation of signal through the cascaded stages of the imaging system (Cunningham 1994, Maidment et al 1993, Siewerdsen and Antonuk 1998, Siewerdsen et al 1998, Neitzel et al 2001), the photoelectrons $S_e$ collected by the CCD pixel can be estimated from the incident photon flux on the detector and the total system gain. $S_e$ represents the photoelectrons generated by the primary photons incident on the detector. As slot scanning eliminates more than 95% of the scattered photons, the contribution of scattered photons is ignored for this application:

$$S_e = QE_s \times CE \times DetGain \times QE_{CCD} \times A_P \int_{E_{\min}}^{E_{\max}} q_{Slot}(E) e^{-\mu_B x} dE,$$

where $q_{Slot}(E)$ is modified energy spectrum for a slot-scanning beam incident on the detector face; $\mu_B$ is the total linear attenuation coefficient of breast tissue; $QE_{CCD}$ is the CCD quantum efficiency and $E_{\min}$ and $E_{\max}$ are the minimum and the maximum photon energies in the x-ray spectrum.

Finally, the pixel intensity in digital units (DU) can be estimated from the total detector signal and the analogue-to-digital (ADC) conversion gain ADCgain:

$$\text{PixInt} = \frac{S_e \times \text{ADCgain}}{S_D} + S_D,$$

where $S_D$ is the added signal in DU estimated from the CCD and ADC data sheets or measured using minimum exposure technique settings.

2.1.1. Energy spectrum simulation. The energy spectrum of the slot-scanning beam is characterized by the x-ray tube settings (tube voltage kVp and tube current mA), the effective beam width ($B_W$) and the scanning speed ($S_{\text{Speed}}$). The photon fluence in photons mm$^{-2}$ modified for the slot-scanning beam for a given tube current and effective scanning time (mAs) can be written as

$$q_{Slot}(E) = q_0(E) mA B_W S_{\text{Speed}},$$

where $q_0(E)$ is photon fluence, which has units of photons mm$^{-2}$ per mAs, taken from published data (Boone 1997) and calibrated for the slot-scanning beam using a measured exposure at a distance of 60 cm from the source and at known tube voltage, tube current, beam width and scanning speed (30 kVp, 100 mA, 5 mm and 22.5 mm s$^{-1}$). Figure 2 shows the calculated spectrum for the given technique factors.

The exposure in ($\mu$Gy) at the surface of the phantom can be estimated from the calculated photon fluence in photons mm$^{-2}$ modified for the slot-scanning beam using the empirically derived equation (Dobbins 2000):

$$E_{\text{Expo}} = \int_{E_{\min}}^{E_{\max}} q_{Slot}(E) [1 - e^{-\mu_B x}] dE \times x(E) \times 8.76.$$

The factor of 8.76 converts mR to $\mu$Gy; $x(E)$ is conversion factor in units of milli-Roentgen (mR) per photons mm$^{-2}$ which is estimated using the empirical equation (Dobbins 2000):

$$x(E) = \left[ x + y \sqrt{E \ln(E)} + \frac{z}{E^2} \right]^{-1},$$
Mathematical tissue-equivalent breast phantom for digital mammography

Figure 2. The tungsten energy spectrum at a distance of 60 cm, tube voltage of 30 kVp, tube current of 100 mA and external filtration of 0.5 mm Al, calculated using equation (4).

Figure 3. The estimated and measured entrance exposure at a distance of 60 cm from the source at different tube voltages, with a correlation of 0.9999, and a slope of 0.96 for 35 kVp. Where x, y and z are constants equal to $-5.023290717769674 \times 10^{-6}$, $1.810595449064631 \times 10^{-7}$ and $0.008838658459816926$, respectively. Figure 3 shows the estimated air kerma exposure for different tube voltages and tube currents using equation (5) compared to the values measured using a relatively large (60 cm$^3$) flat ionization chamber (Radcal 10X5-60E with Radcal 9010 dosimeter, Radcal Corporation, Monrovia). The ionization chamber was placed at a distance of 60 cm from the source. Dose was then corrected for a source-to-detector distance for linear slot scanning, following a $1/r$ rather than a $1/r^2$ law: the field area and hence the flux remains constant in the scanning direction, but decreases in the slot direction and hence dose is proportional to $1/r$, as explained in detail in deVilliers and deJager (2003) and Irving et al.
This relation is confirmed experimentally for linear slot-scanning mammography as shown in figure 4. The calculated exposure using the inverse distance shows a good agreement (with <5% error) for the distance range between 50 and 65 cm from the source where the breast is normally placed in mammography.

The detector exposure, \( D_{\text{Expo}} \), was estimated using the following equation:

\[
D_{\text{Expo}} = \frac{\text{SPD}}{\text{SDD}} \int_{E_{\text{min}}}^{E_{\text{max}}} q_{\text{Slot}}(E) [1 - e^{-\mu_B x}] dE \times \xi(E) \times 8.76, \tag{7}
\]

where SDD is the source–detector distance and SPD is the source–phantom distance. The estimated and the measured exposure show a good correlation of 0.9999, and a slope of nearly 1 for a tube voltage of 35 kVp as indicated in figure 3.

2.2. Estimation of the image contrast

The contrast of an object in a mammography image depends on the density of the object, the total attenuation by the object and the soft tissue in the breast, the x-ray spectrum (photon energy and tube voltage kVp), scattered radiation and the display conditions. The object contrast can be estimated from the difference in the attenuation coefficients of the object and the surrounding background using Dobbins’s relationship (Dobbins 2000):

\[
C = \frac{1}{1 + \text{SPR}} \left[ 1 - \frac{\int q_{\text{Slot}}(E) e^{-\mu_B (D-d) - \mu_A d} \eta(E) dE}{\int q_{\text{Slot}}(E) e^{-\mu_B D} \eta(E) dE} \right], \tag{8}
\]

where \( q_{\text{Slot}}(E) \) is the photon fluence of the slot beam; \( \eta(E) \) is the energy absorption of the scintillator; SPR is the scatter-to-primary ratio, which is disregarded here, as slot scanning eliminates more than 95% of the scattered photons; \( \mu_A \) is the linear attenuation of target tissue and \( \mu_B \) is the linear attenuation of background tissue; \( D \) is the thickness of the background area and \( d \) is the target thickness.

The image contrast of a digital system is defined as the relative difference in intensity between the target tissue (e.g. 100% glandular tissue) and the surrounding region (e.g. 50:50...
glandular/adipose tissue) as shown in equation (9). In a digital system the image contrast is equivalent to the object contrast:

\[ C_{\text{img}} = \frac{S_{\text{background}} - S_{\text{target}}}{S_{\text{background}}}, \quad (9) \]

where \( S_{\text{target}} \) is the mean intensity value in the target area and \( S_{\text{background}} \) is the mean intensity value in the surrounding background.

### 2.3. Modeling the detector noise

Due to the natural fluctuation of the incident photon flux, the electrons generated by the detector obey Poisson statistics. The self-generated noise (quantum noise) and the added noise can then be estimated by using the Poisson and Gaussian distributions. The pixel intensity of a generated image must correspond to the number of incident photons per pixel and the randomly distributed noise. A noisy phantom image can be modeled as follows, with the standard deviation \( \sigma \) representing the total amount of detector noise, while \( n(x, y) \) is a matrix of \( x \) rows and \( y \) columns of random numbers corresponding to image size, which represent a Gaussian distribution with \( \sigma = 1 \):

\[ Ph_n = M_{\text{val}} + \sigma n(x, y), \quad (10) \]

where \( Ph_n \) is the noisy phantom image, while \( M_{\text{val}} \) is the mean intensity pixel value corresponding to the electrons generated by the detector. The standard deviation, which represents the fluctuation of pixel intensity due to the quantum and added noise, can be estimated from the measurement of an image without x-rays (shutter closed) or from a sequence of images of added aluminum sheets using the minimum exposure technique settings available in the machine.

There are several sources of noise that affect the contrast of the phantom image. The detector noise can be estimated from the following:

- **Quantum noise** \( (\sigma_Q) \) due to a natural fluctuation of the photon flux can be estimated by the square root of the generated detector signal:
  \[ \sigma_Q = \sqrt{S_e}. \quad (11) \]

- **Dark noise** \( (\sigma_D) \) is the noise associated with dark current in the CCD and is given by
  \[ \sigma_D = \sqrt{D_S} \quad (12) \]

- **Electronic noise** \( (\sigma_E) \) is all other electronic noise including the amplifier noise.
- **Digitization noise** \( (\sigma_{\text{Dig}}) \) depends on the analogue-to-digital converter (ADC) conversion gain; if the gain is greater than unity, the digitization noise will be considered as a source of noise. The total detector noise expressed in variance is given by
  \[ \sigma_T^2 = \sigma_Q^2 + \sigma_D^2 + \sigma_E^2 + \sigma_{\text{Dig}}^2 \quad (13) \]

### 2.4. Modeling the total system resolution

The amount of blurring or degradation of the phantom image is characterized by the distortion operator, known as the point spread function (PSF), which is convolved with an unblurred image to provide a blurred image. The total distortion of x-ray imaging detectors is characterized by the modulation transfer function (MTF) of the system, which is the Fourier transform of the point spread function, and represents the distortion in various parts of the...
Figure 5. The effect of phantom image blur simulated for different distortion operators (Gaussian filters) on the line pair tools included in the mathematical phantom: (a), (b) and (c) indicate the Fourier transforms of the applied distortion operators and (d), (e) and (f) indicate the amount of blurring from each operator.

detector such as the focal spot MTF, the scintillator MTF, pixel blurring and the blurring due to the mismatch between the scanning velocity and the transfer rate of charge in the CCD.

Figure 5 shows the Fourier transforms of applied distortion operators, which represent the estimated MTF of the imaging system, and their effect. The line pair tools are used to visualize the amount of phantom image blurring in response to the application of different distortion operators. The system MTF in figure 5 characterizes the total distortion from all the sources mentioned above.

2.5. Generation of phantom image

A grayscale intensity image is generated which contains the body of the phantom (50:50 glandular/adipose tissue) and areas representing the other structures contained in the phantom as described earlier. The intensity of pixels within the phantom body or in each area is defined by the attenuation of each structure (as a function of its density, the linear attenuation coefficient and the thickness) or by its contrast against the background.

The phantom image parameters such as the pixel intensity value, the location of structures within the phantom image and the size and dimensions of the phantom are stored in a phantom parameter matrix. The phantom image can be modeled as follows.
• An initial phantom image is generated consisting of x rows and y columns with a default pixel intensity value equal to the intensity value corresponding to the digital value of the 50:50 glandular/adipose tissue signal.
• The pixel intensity value in digital units (DU) is then modified for each structure according to its contrast:

\[ P_h(x, y) = P_{h_{in}}(x, y) + A, \]

where \( P_{h_{in}}(x, y) \) is the initial pixel intensity value and \( A \) is added intensity for each structure.
• Quantum noise due to the nature of the x-rays is then added to the generated phantom image.
• The phantom image is then modulated using the system resolution (MTF) as a distortion factor.

Figure 6 shows a phantom image generated using a tube voltage of 30 kVp, tube current of 200 mA and 4000 electrons of additive noise, which approximately correspond to the measured variance using a dark image (no x-ray) (discussed in section 3.1).

3. Validation of the mathematical phantom

The mathematical breast phantom was validated using a series of digital images of a physical tissue-equivalent phantom designed for mammography and manufactured by Computerized Imagining Reference Systems (CIRS), USA, on which the mathematical phantom was based. Phantom images were acquired using the prototype slot-scanning mammography system developed at MRC/UCT Medical Imaging Research Unit, University of Cape Town. The physical phantom was used to measure the pixel intensity, subject contrast, contrast-to-noise ratio (CNR), analogue-to-digital converter (ADC) conversion gain and detector additive noise. These measured values were used to validate the simulation of mathematical phantom images and the estimation of the detector noise and the conversion gain.

3.1. Estimation of digital pixel intensity and ADC conversion gain

The number of photoelectrons per pixel generated on the detector was estimated for different technique factors (tube voltage: 25, 30, 35 and 40 kVp; tube current: 16–200 mA) using
Figure 7. Least-squares fit of the measured pixel intensity and the estimated pixel signal in electrons calculated from the measured detector exposures of 30 kVp.

Table 2. Parameters used in obtaining digital images of the physical phantom.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tube voltage</td>
<td>25, 30, 35 and 40 kVp</td>
</tr>
<tr>
<td>Tube current</td>
<td>16–200 mA</td>
</tr>
<tr>
<td>Scanning speed</td>
<td>22.5 mm s⁻¹</td>
</tr>
<tr>
<td>CCD pixel binning</td>
<td>1 × 1</td>
</tr>
<tr>
<td>Source-to-phantom distance</td>
<td>60 cm</td>
</tr>
<tr>
<td>Collimator gap</td>
<td>0.8 mm</td>
</tr>
<tr>
<td>Pixel size</td>
<td>0.048 mm</td>
</tr>
</tbody>
</table>

equation (2). A series of digital images of the physical phantom were acquired at different technique factors as shown in table 2.

The mean pixel value in 50:50 glandular/adipose tissue in the phantom was measured for each digital image in a region of interest comprising a 100 × 100 pixel area. The pixel values for a tube voltage of 30 kVp at different tube currents were plotted against the number of estimated photoelectrons for each detector exposure (different mA) as shown in figure 7.

Using a least-squares fit of the measured pixel intensities in digital units (DU) and the estimated detector signal in (e⁻), the digital pixel values at different exposures can be estimated using the following equation:

\[
DU = 2.0298 \times e^- + 9851. 
\]  

The sensitivity of the detector is given by the slope (2.0298 DU/e⁻) of the fitted line in figure 7, which is slightly less than the value (2.0 e⁻/DU) set manually by the operator. The conversion gain is given by the reciprocal of the slope and was found to be approximately equal to 0.5 e⁻/DU.

The additive detector signal (dark signal) can also be estimated for the fitted line and was found to be 9851 digital units (DU), which is approximately equivalent to 4000 electrons.
value agreed (within 95%) with the measured pixel intensity for an image taken without x-ray exposure. Figure 8 shows the pixel intensity estimated, using equation (15), for tube voltages 30, 35 and 40 kVp, compared to the measured values. The error bars in figure 8 indicate the range of the measured and simulated pixel intensity for a selected region of interest (i.e.
the measured pixel intensity for each selected region is averaged over three readings and the simulated intensity is averaged over three simulations).

Figure 9 shows the difference between measured and simulated pixel intensity values as a function of their mean, as suggested by Bland and Altman (1986). The simulated pixel intensity values are calibrated using the measured values at 30 kVp; this is reflected in the graph, which shows small differences at 30 kVp, but differences up to 4% and 6% for 35 kVp and 40 kVp, respectively.

3.2. Estimation of the image contrast

Figure 10 shows the estimated and the measured contrast for different glandular tissues in the phantom as a function of tube voltage (kVp) using equations (8) and (9).

A t-test was performed to compare the measured and simulated contrast values, and no statistically significant difference was found ($p > 0.05$ in all cases).

3.3. Estimation of contrast-to-noise ratio

The contrast-to-noise ratio is the relative difference in signal between a target and the surrounding background divided by the inherent noise in the image (equation (16)). Noise usually is the measured standard deviation $\sigma$ of a region of interest in the background:

$$\text{CNR} = \frac{S_{\text{target}} - S_{\text{background}}}{\sigma} = \frac{S_{\text{target}} - S_{\text{background}}}{\sigma}. \quad (16)$$

Figure 11 shows a visual comparison between the estimated and the measured image contrast for different tissue-equivalent regions included in the mathematical phantom at different tube voltages and tube currents. Figures 12 and 13 show the simulated contrast-to-noise ratio (CNR) for tube voltages 30 and 40 kVp compared to the measured CNR from the digital images of the physical phantom. The error bars in figures 11 and 12 indicate the range in CNR in each region of interest (three readings were taken).

Figure 14 shows the difference between the measured and simulated CNR as a percentage of their mean. Estimated CNR values depend on the estimation of pixel intensity and noise; hence higher differences are expected. Differences up to 22% are shown. The actual density and composition of the component regions of the physical phantom may play a role in...
Figure 10. The mean estimated and measured image contrast of different structures in the phantom as a function of tube voltage: (a) 100% glandular tissue; (b) 30:70 glandular/adipose; (c) 70:30 glandular/adipose and (d) 100% adipose tissue. The error bars indicate the range in the image contrast, as three image contrast readings were taken at each setting.

producing larger differences, if the theoretical values used in the simulations misrepresent the physical values. The differences between the measured and simulated CNR are more pronounced in the 30% and 70% glandular tissue, which may indicate disagreement between the theoretical and physical composition of these regions. Our larger differences occur at lower tube current, similarly to those of Doyle et al. (2006), who reported differences in the measured and estimated CNR for chest radiography of up to approximately 20% as acceptable.

The noise texture is different in the physical and simulated images; the simulated images show higher apparent granularity, which may affect lesion detectability in them.

4. Phantom application: beam optimization

There are limitations in using a polychromatic x-ray spectrum, i.e. the presence of a wide range of x-ray energies, in mammography. Firstly, the low-energy photons are mainly absorbed by the breast tissue and serve to increase the absorbed dose to the patient. Secondly, the higher energy photons increase the contribution of Compton scatter, which degrades the image quality. A narrow energy spectrum can provide better contrast and significant dose reduction (Burattini et al. 1995, Boone and Siebert 1994, Lawaczeck et al. 2005). The initial evaluation of a prototype digital mammography unit operating with a nearly monochromatic x-ray beam showed that the use of monochromatic slot-scan mammography resulted in correct identification of 93% of calcifications within a contrast-detail phantom (Diekmann et al. 2004). Clinical installation of
monochromatic x-ray sources for mammography would require high-power x-ray tubes and imaging with the slot-scanning technique (Lawaczeck et al 2005).

The mathematical breast phantom was used to study the feasibility of using a monochromatic filter MXF (MXF 2006) for spectrum optimization in slot-scanning mammography. This filter produces a narrow monochromatic beam from a polychromatic x-ray beam using Bragg’s law. The monochromatic beam was simulated using a Gaussian distribution according to the data provided by the MXF manufacturer (MXF 2006). Figure 15 shows the calculated monochromatic beams for different tuned energies. The detective quantum efficiency (DQE) was estimated using cascaded linear analysis (Siewerdsen and Jaffray 2000). Mean glandular dose (MGD) was calculated using Boone’s fit equations (Boone 2002).

Figure 16 shows the effect of using the MXF monochromatic beam on detector performance in terms of image contrast and dose reduction. The bar plots in figure 16 above each pair of images show the ratios of monochromatic to polychromatic performance (image contrast, mean glandular dose (MGD), contrast-to-noise ratio (CNR) and system detection efficiency (DQE)); values above unity represent an improvement in the image quality parameters and an increase in the mean glandular dose. A monochromatic x-ray beam with a monochromatic filter tuned to 19 keV will result in a 60% improvement in contrast as shown in the bar plots in figures 16(a)–(c). Using the narrow spectrum will also result in dose reduction and a consequent reduction in photon flux. To achieve the same contrast-to-noise ratio using the two types of beams, an increase in scanning time or tube loading is required for the monochromatic beam. For the 19 keV monochromatic beam (10% FWHM), tripling the scanning time will result in the same contrast-to-noise ratio with a 50% dose reduction with respect to the polychromatic beam as shown in figure 16(c). Using a wider monochromatic
beam (e.g., 25% FWHM) produces similar improvements as tripling the scanning time of the 10 FWHM monochromatic beam, as shown in figure 16(d).

5. Discussion

A breast tissue-equivalent mathematical phantom was simulated for digital mammography to resemble a physical tissue-equivalent phantom designed for mammography. The exposures estimated for different tube voltages and tube currents showed good agreement with the measured values (with good correlation and slope of nearly 1, as indicated in figure 3). The estimated exposure using the inverse distance ($1/r$) rather than inverse square distance ($1/r^2$) shows good agreement (with $< 5\%$ error) for the distance range between 50 and 65 cm from the source where the breast is normally placed in mammography. These estimated
Figure 13. Measured and simulated contrast-to-noise ratio (CNR) for different tissues in the phantom image at 40 kVp tube voltage: (a) 100% glandular tissue, (b) 70% glandular tissue, (c) 30% glandular tissue and (d) 100% adipose tissue. The error bars indicate the range in CNR in each region of interest.

exposures were used for calibration of the incident energy spectrum for a given technique factor. The estimated analogue-to-digital conversion gain shows good agreement with values set manually. The additive noise due to dark signal and electronics was measured using digital images acquired with x-ray radiation off. The image contrast and the contrast-to-noise ratio (CNR) for different regions simulating the glandular and the adipose tissue showed good agreement with the measured values.

The agreement between the measured physical images and the measured image quality parameters (image contrast and CNR) with the estimated values allows the use of this mathematical phantom for parameter optimization in slot-scanning mammography. Caution should be exercised at low tube currents as these produce higher simulation errors. In addition, the composition of the physical phantom as a source of difference between measured and simulated values should be investigated.

The optimization of the x-ray spectrum using a monochromatic filter results in better contrast and the same image quality (CNR and DQE), with a significant dose reduction. The
Figure 14. Differences between the measured and simulated CNR for 30 kVp and 40 kVp as a percentage of their mean.

Figure 15. The estimated MXF monochromatic beams for different tuned energy levels and the unfiltered spectrum.

performance of the monochromatic filter depends on selection of spectrum tuned energy and the beam width (FWHM).

It should be noted that the simulated phantom presents a uniform background rather than the randomly structured background of breast tissue, that the sizes of the objects simulating lesions may be insufficient for the task and that sub-pixel alignment has not been addressed. For these reasons, the simulation model may not be sufficient for the evaluation of lesion detectability.
Figure 16. The image performance of the polychromatic (left) and the monochromatic (right) beams at tube voltage (30 kVp) and tube current (100 mA)—the polychromatic beam is the unfiltered spectrum shown in figure 15 and the monochromatic beam is tuned to 19 keV: (a) performance using the same technique factors; (b) doubling the scanning time of the monochromatic beam; (c) tripling the scanning time of the monochromatic beam and (d) doubling the scanning time and using FWHM of 25% rather than 10% for the monochromatic beam.

The scope of the model presented is limited to the simulation of the CIRS-based phantom imaged with a slot-scanning system. Therefore, the model, and the results of simulating a monochromatic beam, may not be generalizable beyond the specified system.
For slot-scanning digital mammography, the model estimates both qualitative and quantitative image quality parameters for detector design and optimization, and provides both quantitative and visual assessment of system performance. Simulation of structures in the mathematical phantom that exist in a physical phantom designed for mammography allows direct comparison between simulated and acquired images for a given system and allows calibration of the mathematical phantom. The model also provides visual tools for the estimation of system resolution, which is limited by the image matrix size (maximum of 7 lp mm$^{-1}$ for 12 bit image, $4069 \times 4069$ pixels).

The model provides simulation tools which can be easily calibrated for different x-ray tubes and detectors. Most of the available simulated breast phantoms (Bakic et al 2002,
Bliznakova et al 2003, Hunt et al 2005, Dance et al 2005, Reiser et al 2006) are modeled to resemble anatomical structures and breast composition realistically, and to allow for the simulation of anatomical variations. These phantoms cannot be easily used for mammography signal calibration and quality control. Most of the quality control phantoms are physical phantoms, which need physical measurement for system calibration and optimization. An easily calibrated software quality control phantom will be useful in detector design and optimization.

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