The X-ray detector is the heart of a digital mammography system. Its improved characteristics of dynamic range and signal-to-noise ratio provide inherent advantages over screen-film technology. Detector technologies used for digital mammography can be distinguished by the acquisition geometry into scanning or full-field detectors, by energy conversion mechanism into phosphor-based and nonphosphor-based detectors and by how the detector signal is converted into an image value into signal-integrating and quantum-counting systems. Reading of the detector signal can be integrated into the detector assembly or the detector can be in the form of a sensitive plate in a portable cassette, which is moved to a separate device for readout. Detector performance characteristics vary among these different technological approaches. An understanding of the physics on which detector operation is based can help explain these differences. Various image processing operations can be carried out to correct for spatial nonuniformities in detector response and to improve the effective spatial resolution of the detector. In addition, use of a digital detector provides opportunities for more sophisticated automatic control of exposure factors for image acquisition.

The detector is one of the defining features of a digital mammography system. The detector produces an electronic signal that represents the spatial pattern of
X-rays transmitted by the breast. The detector is designed to overcome several of the limitations inherent in the screen-film image receptor used in analog mammography, and in so doing, potentially provides improved diagnostic image quality and a reduction of dose to the breast.

The function of the detector can be described by a set of sequential operations that include:

(a) Interaction with the X-rays transmitted by the breast and absorption of the energy carried by the X-rays
(b) Conversion of this energy to a usable signal—generally light or electronic charge
(c) Collection of this signal
(d) Conversion of light to electronic charge (in the case of phosphor-based detectors)
(e) Readout of charge, amplification, and digitization

These operations must be optimized if the detector is to provide high-quality images at appropriate dose levels. Detectors are characterized by their quantum detection efficiency, sensitivity, spatial resolution properties, noise, dynamic range, and linearity of response.

As discussed in Chap. 1, digital images are acquired by sampling the pattern of X-rays transmitted by the breast. In practice, this is often accomplished using a detector that is constructed as an array of discrete detector elements or dels, each of which more or less independently measures the X-rays incident on it. The pitch or spacing between dels and the dimensions of the active portion of each del (aperture) in part determine the spatial resolution properties of the imaging system. The concept of spatial resolution and its quantification in terms of the modulation transfer function (MTF) were introduced in Chap. 1.

2.2 Geometric Considerations

A detector that is suitable for mammography must be able to capture the transmitted X-ray pattern from as much of the breast as possible. To satisfy this requirement, it must have spatial dimensions of approximately \(24 \times 30 \text{ cm}\). Positioning of a small breast can be facilitated if a smaller detector size (format of approximately \(18 \times 24 \text{ cm}\)) is also available. Alternatively, the imaging system can be designed such that when a small breast is imaged on \(24 \times 30 \text{ cm}\) detector, the collimated X-ray beam and the detector area over which the image is recorded can be shifted to cover the appropriate tissue in the axilla, depending on whether the left or right breast is being imaged.

It is important to be able to image the breast as close to the chest wall as possible. Therefore, there should be the minimum possible amount of insensitive material associated with the detector at this edge of the image. Finally, an excessively thick detector assembly can impede positioning of the breast for some views.

Two major types of digital mammography systems have been introduced. One uses a detector that is the full size of the field that is to be imaged. The other employs a detector array that is long in one dimension and narrow in the other. In the first type, “snapshot” imaging is performed by acquiring the X-ray transmission information from all parts of the breast simultaneously. As in film-based imaging, a radiographic grid is generally used in such systems to reduce the loss of quality caused by recording scattered X-rays.

In the second type of imager, the detector is scanned across the breast in synchrony with a long, narrow collimated X-ray beam to acquire the image progressively. Because the X-ray field is smaller, there is less scattered radiation recorded and these systems are operated without a radiographic grid.

2.3 Basic Physics of X-Ray Detectors

The initial stage in the detector, X-ray interaction is common to all detector technologies. This occurs at the level of individual atoms of the detector material. At the energies used for mammography, X-rays incident on the detector interact by one of the three mechanisms, elastic scattering, inelastic (Compton) scattering, or the photoelectric effect. Elastic scattering leaves no energy in the detector and produces no signal. Compton scattering results in part of the energy of the incoming X-ray being absorbed at the initial point of X-ray impact liberating an energetic recoil electron, but the remainder is carried away by the scattered quantum to be deposited elsewhere, resulting in the loss of spatial resolution. In a photoelectric interaction, the incoming X-ray knocks an electron out of an inner (K-shell or L-shell) orbital of the atom (Fig. 2.1) and much of the energy of the X-ray is transferred to this “photoelectron.” When the vacancy is refilled by an electron from a more loosely bound shell, the remainder of the energy is
Detectors for Digital Mammography

transferred either to a second (Auger) electron or to a relatively low-energy fluorescent X-ray (which will be absorbed to liberate another electron). Therefore, when X-rays interact and lose energy through either the Compton or photoelectric mechanisms, this appears as kinetic energy of electrons. Because these are charged particles, they interact intensively with nearby atoms and their kinetic energy is lost within a short distance (∼1 mm) from the initial interaction site. This is desirable because it limits the blurring that would result from spreading of the energy and allows high spatial resolution to be achieved. Therefore, it is desirable that most interactions in the detector are of the photoelectric type and this can be achieved by employing a detector material having a relatively high atomic number. For example, for iodine and selenium, at 20 keV, 94% and 96%, respectively, of X-ray interactions will be by the photoelectric effect.

For all detector types, the transfer of energy to the detector material as the photoelectrons and recoil electrons slow down occurs through excitation or ionization of electrons in neighboring atoms within the detector structure. What occurs next depends on the specific molecular structure of the detector. We can distinguish between structures such as noble gases, photoconductors, fluorescent phosphors, and photostimulable phosphors.

In solids such as crystals and some amorphous materials, the orbitals associated with individual atoms become less apparent and the solid behaves as if the electrons (especially those in the outermost shells) exist in continuous bands, each occupying a range of energy levels, separated by unpopulated forbidden regions (bandgaps) as illustrated in Fig. 2.2. The electrons normally reside in the so-called “valence band.” In nonmetallic solids, this band is separated by a bandgap of energy $E_G$, from the “conduction band” in which the electrons are essentially free.

2.3.1 Photoconductors

In photoconductors, excitation (B in Fig. 2.3) of each electron in the molecular lattice to the conduction band results in a free electron and a “hole” left behind in the valence band. Because there are competing mechanisms of energy transfer, on average an energy $w ∼ 3E_G$ is required to produce each electron–hole pair. Under the influence of an electric field, the electrons will drift in one direction in the conduction band and the holes in the other in the valence band, resulting in a movement of charge that is proportional to the energy deposited in the material by the X-rays (Fig. 2.3).


2.3.2 Phosphors

Phosphors are crystalline materials to which impurities have been added to create energy levels within the forbidden bandgap. Electrons excited into the conduction band (B in Fig. 2.4a) are captured by these impurity centers and rapidly de-excite (typically in nanoseconds), passing through these energy levels on their way down to their ground state (valence band). The energy levels of the impurities are chosen so that in transitioning between these levels, the electron causes energy to be emitted in the form of light (D). The energy of these light quanta is equal to the difference in energy levels, \( \Delta E \), for the transition and this determines the color (wavelength, \( \lambda \)) of the light by \( \lambda = \frac{hc}{\Delta E} \), where \( h \) is Planck’s constant and \( c \) is the speed of light in vacuum.

2.3.3 Photostimulable Phosphors

Photostimulable phosphors (Fig. 2.4b) are phosphors that contain a large number of electron trapping sites, also called f-centers or color-centers, located in the forbidden bandgap. Electrons excited to the conduction band have approximately equal probabilities of either producing light or being trapped at one of these sites, interrupting their de-excitation. Electrons can remain trapped for minutes to days until they are detrapped by exposure to a stimulating light (E). The stimulating energy is sufficient to cause them to be re-excited to the conduction band, where again they have a probability of de-excitation to the valence band, producing light (D) as in a conventional phosphor.

2.3.4 Noble Gases

In noble gases, energetic photoelectrons and recoil electrons cause electrons in outer atomic orbitals to be freed completely from the atom (ionized), resulting, for example in free electrons and positively charged gas ions (Fig. 2.5). These charged entities can be collected in an electric field to form a charge signal proportional to the amount of absorbed X-ray energy.

2.4 Aspects of Detector Performance

2.4.1 Quantum Detection Efficiency

As discussed in Chap. 1, quantum detection efficiency, \( \eta(E) \), describes the fraction of the X-rays falling on the detector, that interact with it, producing at least some signal. Some calculated values of \( \eta \) for
Detectors for Digital Mammography

17

different detector materials are given in Fig. 2.6. In general, $\mu$ decreases as energy increases, causing $\eta$ to do so as well. An exception occurs when the X-ray energy exceeds the absorption energy threshold for an atomic orbital (an “absorption edge”) of the detector material. For example, as seen in Fig. 2.6 for the CsI phosphor, $\eta$ increases dramatically at 33 keV because of the sudden increase in the attenuation coefficient of iodine at its K absorption edge. Some data on detector materials used for digital and film mammography are given in Table 2.1.

2.4.2 Sensitivity

Referring back to (1.2) in Chap. 1, for a certain mean number of X-rays, $n_d$, incident on the detector, the number, $n_a$, interacting with the detector will be:

$$n_a = n_d \eta(E) \quad (2.1)$$

The number of secondary light quanta or electrons produced will be:

$$n_{sq} = n_a g \quad (2.2)$$

Here, $g$ is the gain. For the initial energy conversion in the detector (depending on detector design, more than one conversion stage may exist), $g = E_{abs}/w$, where $E_{abs}$ is the energy absorbed in the detector per interacting X-ray and $w$ is the amount of energy required to produce an element of signal (a light quantum or an electron, whichever is being measured). Here, the expression, $E_{abs}$, is used to reflect the fact that part of the energy, $E$, carried by the X-ray may not be absorbed due to escape of scatter and X-ray fluorescence.

The sensitivity of the detector, therefore, depends on: (a) $\eta(E)$, (b) $g$, (c) the efficiency of signal collection and measurement of the charge that is produced. Note that both $\eta$ and $g$ for the initial conversion stage, depend on $E$.

The X-ray interaction layer of the detector can be fabricated from materials with relatively high atomic numbers ($Z > 30$). At the low energies typically used in digital mammography (<50 keV), the majority of X-ray interactions in the detector will be through the photoelectric effect. The photoelectron will lose its energy within a very short distance of the point of initial interaction so that there will be little spread of charge. For example, for the cesium in CsI(Tl) phosphor, over 90% of interacting 30 keV X-rays are absorbed through the photoelectric effect. The remainder will interact through scattering, but there is a high probability that their energy will subsequently be absorbed locally. For incident X-rays above the K edge of the detector material, some of

![Fig. 2.6. Quantum interaction efficiencies of detector materials used in mammography for 0.1 mm thickness (redrawn from YAFFE and ROWLANDS 1997. With permission from IOP)](image)

Table 2.1. Characteristics of some common detector materials

<table>
<thead>
<tr>
<th>Material</th>
<th>$Z$</th>
<th>$E_K$ (keV)</th>
<th>$W$ (eV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>CdTe</td>
<td>48/52</td>
<td>26.7/31.8</td>
<td>4.4</td>
</tr>
<tr>
<td>High purity Si</td>
<td>14</td>
<td>1.8</td>
<td>3.6</td>
</tr>
<tr>
<td>Amorphous selenium</td>
<td>34</td>
<td>12.7</td>
<td>50 (at 10V/μm)</td>
</tr>
<tr>
<td>CsI(Tl)</td>
<td>55/53</td>
<td>36.0/33.2</td>
<td>19</td>
</tr>
<tr>
<td>Gd$_2$O$_2$S</td>
<td>64</td>
<td>50.2</td>
<td>13</td>
</tr>
<tr>
<td>BaFBr:Eu (as photostim. phosphor)</td>
<td>56/35</td>
<td>37.4/13.5</td>
<td>50–100</td>
</tr>
<tr>
<td>Xenon gas</td>
<td>54</td>
<td>34.6</td>
<td></td>
</tr>
</tbody>
</table>
the energy will be re-emitted as X-ray fluorescence, although again, it may be reabsorbed, particularly if careful attention is paid to detector design, for example incorporating a choice of materials appropriate for the energy spectrum used for imaging and ensuring that one type of atom absorbs the fluorescence produced by another. In this way, it is possible to create a detector with \( E_{\text{abs}} = E \).

For a particular X-ray energy, the amount of signal produced on interaction will be inversely related to the value of \( w \) of the detector material. Values of \( w \) (in electron volts) are given for various detector materials in Table 2.1. An X-ray quantum of energy 25 keV (20,000 eV) can potentially produce 25,000/50 = 500 electron–hole pairs in a selenium detector. From Fig. 2.6, a 0.1-mm thick Se detector will have \( \eta \sim 90\% \) at 20 keV, so the sensitivity is 450 electron–hole pairs per incident X-ray.

### 2.4.3 Noise in Detectors

In Chap. 1, the basic concepts of noise in X-ray imaging were introduced and the dependence of one source, referred to as X-ray quantum noise, on the number of X-rays used to form the image, was discussed. In addition, several quantities used to describe imaging performance, namely the signal-to-noise ratio (SNR), the signal-difference-to-noise ratio (SDNR), the detective quantum efficiency (DQE), and the noise equivalent quanta (NEQ) were reviewed.

In an ideal imaging system, the only source of noise in the image should be quantum noise. Quantum noise is fundamental and unavoidable in an X-ray image, but for a given exposure, its effect should be minimized by ensuring that the quantum efficiency of the detector is as close to 100% as possible. In real imaging systems, there are also other noise sources, mainly associated with the detector, and efforts should be made in the system design to ensure that these are much smaller than the quantum noise.

One form of noise is the structural fluctuation in sensitivity over the area of the detector. In screen-film mammography, this effect cannot be removed, only minimized through design of the screens and film and tight quality control standards in manufacturing. In digital mammography, where the same detector is used repeatedly, if these del to del sensitivity differences remain constant, over time they can be considered as fixed pattern noise. Because the image is captured in digital form, the effects of fixed pattern noise can largely be removed in most digital detector systems. A flat-fielding or gain correction is applied to each acquired image.

All detectors convert the X-ray energy into a secondary signal such as light in a phosphor or electronic charge in a direct conversion type detector. These processes also introduce statistical fluctuation, i.e., noise over and above that caused by the primary quantum noise.

The fluctuation or noise in the number of second- ary quanta from which the image signal is derived depends on both the Poisson fluctuation in \( n_a \) and in the fluctuation, \( \sigma_g \) in \( g \). For a system with a single conversion stage such as that described by (2.2):

\[
\sigma_{sq}^2 = \sigma_{sq}^2 g^2 + n_a \sigma_g^2, \tag{2.3}
\]

where the first term represents the quantum noise and the second the additional noise caused by fluctuation in the gain. The noise is then written as:

\[
\sigma_{sq} = \sqrt{n_a (g^2 + \sigma_g^2)}. \tag{2.4}
\]

There are various factors that can cause fluctuation in the gain. For example, because a polyenergetic spectrum of X-rays is used for imaging the breast, the absorbed energy resulting from an interacting X-ray will depend on the initial energy carried by that particular quantum. Therefore, as quanta of different energy in the spectrum are absorbed, the amount of light or charge produced will fluctuate from quantum to quantum, giving rise to an apparent fluctuation in \( g \).

In addition, even for an absolutely constant amount of energy deposited by an X-ray quantum, the energy will be absorbed through a random chain of different types of subsequent interactions in the detector material and only some of these will give rise to emission of light. This variability will cause additional statistical fluctuation in the gain, a phenomenon described by Swank (1973). These additional sources of noise reduce the SNR for a given X-ray exposure to the detector, thereby reducing DQE.

For example, from (2.2) and (2.4), the SNR for a simple detector can be written as:

\[
\text{SNR} = \frac{n_{\text{tot}}}{\sigma_{n_{\text{tot}}}} = \sqrt{n_a} \frac{1}{\sqrt{1 + \frac{\sigma_g^2}{g^2}}}. \tag{2.5}
\]
Therefore, SNR can be maintained at an appropriate level by ensuring that \( n \) is adequate and also that \( g \) is large when compared with \( \sigma \). Frequently, this condition can be achieved by employing a fairly high level of gain.

Detectors often have multiple gain stages, and multiple stages of energy conversion, some where the gain is less than 1.0, i.e., there is a loss of light quanta or electrons. An example of this is where light from a phosphor screen is collected with a lens to be recorded with an optical detector. This occurs in photostimulable phosphor detectors described below. If the number of secondary signal quanta collected and detected per interacting X-ray is not much larger than one, then we say that a “secondary quantum sink” exists. In this case, the statistical fluctuation in the detection of the secondary quanta becomes a significant noise source and will reduce the SNR and DQE. For this reason, it is important that the gain associated with the detector is large enough to offset any losses due to inefficiency in signal collection.

### 2.5 Detector Corrections

#### 2.5.1 Uniformity Correction

The sensitivity of the imaging system should be spatially uniform so that any variations in the image signal can be attributed to structures in the breast.

In most types of digital mammography systems, an algorithm to correct for nonuniform sensitivity in the imaging system is applied to all images. The procedure is variously referred to as gain correction or “flat-fielding” and is illustrated in Fig. 2.7. If the detector has a linear response to X-rays, the response of each del can be described by a straight line having a slope, which represents its gain and an intercept representing the “dark signal” (detector output in the absence of radiation). Two such dels are illustrated in Fig. 2.7a. To perform the flat fielding correction, the slopes and

![Fig. 2.7. Approach to flat field correction of a digital mammography detector. (a) A system with linear response illustrating two dels with different dark signals (intercepts) and gains (slopes). Enlarged view near the bottom end of the range illustrates dark signals, \( D_1 \) and \( D_2 \). These are measured by acquiring images without radiation. (b) Response has been corrected for different dark signals. Gain is still different. Exposure to a fixed amount of radiation \( E_{cal} \) allows determination of the slopes \( G_1/E_{cal} \) and \( G_2/E_{cal} \) for each del. (c) Response of the dels after correction. (d) Systems with nonlinear response cannot be completely corrected using this simple method (from Digital Mammography, eds. ED Pisano, MJ Yaffe, CM Kuzmiac. Lippincott, Williams and Wilkins, a Walters Kluwer Company, 2004. With permission)
intercepts describing every del in the detector must be measured. This is quite straightforward to do with a digital detector. The first step is that a "dark" image is obtained, by recording the detector response for the time equal to that of an X-ray exposure, but without X-rays. The pixels in this image consist of the intercepts $D_1, D_2, \text{etc.}$ from all dels in the detector as shown in the inset to Fig. 2.7a) and these values are stored. In any subsequent image acquisition, these intercept values are subtracted from the measurement arising from each corresponding del, resulting in an image where it appears that the dark signals from all dels are zero. Such an offset correction can be made as frequently as necessary to compensate for temperature-related detector variations. At this point, it is possible to correct for differences from del to del in the slopes or sensitivities.

This is done by exposing the detector to an X-ray beam that has passed through a uniform attenuator. The constant exposure, $E_{\text{cal}}$, received by all dels will produce different signals according to the sensitivity of each del (Fig. 2.7b). This image, which is essentially a map of sensitivities, is stored and used to correct the response of each del in subsequent images, so that it appears that all dels provide uniform response (Fig. 2.7c).

An example of the effect of a flat-fielding correction is shown in Fig. 2.8. The interval between calibrations to measure flat-fielding correction constants depends on the temporal stability of the detector.

Because of the mask used for flat field correction is an X-ray image, it will contain quantum noise and this will be added to the digital mammogram when the correction is applied. If the digital mammogram and the mask image were produced with the same amounts of radiation, the standard deviation of the image pixels would be increased by $\sim \sqrt{2}$, i.e., about 40%. To avoid unnecessary increase in image noise, it is important that the flat-field mask be produced using a much larger amount of radiation than used for each individual mammogram. This can be most easily accomplished by averaging many acquired mask images together to form the working mask. For example, if the equivalent radiation for ten images were used to form the mask, the noise in the mask would be reduced by $\sqrt{10}$ or about threefold and the increase in noise due to flat-fielding would be $\sqrt{1.1}$ times that of the uncorrected image, i.e., virtually unchanged.

The flat-fielding procedure described above essentially removes all spatial variation in what is assumed to be a uniform imaging field. Nonuniformities in signal that are not due to the detector, but are caused by such phenomena as heel effect of the X-ray tube, variation in X-ray path length through air (inverse square law), the beam filter, compression plate, and

![Fig. 2.8. Digital mammogram (a) before and (b) after flat-field correction (from Digital Mammography, eds. ED Pisano, MJ Yaffe, CM Kuzmiac. Lippincott, Williams & Wilkins, a Walters Kluwer Company, 2004. With permission)]
antiscatter grid will also be completely or partially removed. If the flat fielding calibration is performed under one set of conditions, but imaging is done under another (e.g., a change of kV, target, or filter material, use of a compression plate of a different thickness or composition), the flat fielding procedure may generate artifacts.

The flat-fielding correction is generally based on an assumption that the detector responds linearly to radiation exposure. If any or all of the detectors have nonlinear response, then to obtain the calibration constants used to generate artifacts. Under other signal levels, different nonlinearities of the detector responses will result in image nonuniformities (Fig. 2.7d).

2.5.2 Resolution Restoration

As discussed previously, there is a loss of spatial resolution in most detectors due to lateral spreading of signal. This causes the MTF to be reduced. But because the image is stored in digital format, it is possible to apply a correction to restore at least part of the drop in MTF and cause the image to be sharper. The method of implementing such corrections is proprietary to each manufacturer; however, a generic explanation can be offered. One approach to restoration is to perform a Fourier transformation of the image to represent it in the spatial frequency domain (what in MRI parlance is referred to as K-space). The MTF is a frequency-domain representation of how the imaging performance drops at each spatial frequency. By dividing the Fourier-domain image by the falling MTF, this drop is compensated. Then, by performing an inverse Fourier transform, the image is brought back to real-world coordinates, corrected for the resolution loss. This would be a perfect restoration except that images contain noise and when such a restoration is performed, the noise is amplified. For this reason, a modified restoration is generally performed. The noise level at each spatial frequency is automatically measured and a weighting function is established such that full restoration is applied at low spatial frequencies where the noise is low compared with the image signal and progressively less restoration is applied at higher spatial frequencies at which noise levels are relatively high.

An alternative, but entirely equivalent restoration can be applied without the need for the two Fourier operations by using a process called deconvolution.

2.6 Linear vs. Logarithmic Response

X-rays are attenuated in an exponential manner as they pass through matter. If we could image the breast with monoenergetic X-rays, the number of X-rays, \( n \), that would arrive at the detector having followed a particular straight-line path through the breast would be:

\[
n = n_0 e^{-\sum \mu(z) \Delta z},
\]

where \( n_0 \) is the number of X-rays incident on the breast, \( \mu(z) \) is the attenuation coefficient for a tissue element of size \( \Delta z \) at location \( z \). A detector having a linear response to X-rays produces a signal proportional to the number of X-rays transmitted and, therefore, exponentially related to the actual tissue properties. If instead, the detector signal was proportional to the logarithm of \( n \) instead of \( n \) itself the signal would be:

\[
\log \left( \frac{n_0}{n} \right) = \sum \mu(z) \Delta z.
\]

Such a signal would be more directly related to the tissue composition along the path. Logarithmic transformations can be applied to the detector data and this is done in photostimulable phosphor detector systems, with the effect of reducing the range of signal that must be digitized. Once the signal has been transformed in this way, the simple linear flat-field corrections described above can no longer be performed. Alternatively, logarithmic or similar types of transformations can be applied after flat-fielding, during image display, to compress the range of the data.

2.7 Detector Types

Several different types of detectors are used for digital mammography. These are briefly described here.

2.7.1 Phosphor-Flat Panel

Phosphor flat panel detector systems (Fig. 2.9) are based on a large-area glass plate. Using solid-state manufacturing techniques, a rectangular array of light-sensitive photodiodes is deposited onto the plate. These
are interconnected with an array of control and data lines as well as a thin film transistor (TFT) switch adjacent to each photodiode. These electronic components are fabricated using amorphous silicon technology.

X-rays are absorbed by a layer of thallium-activated cesium iodide phosphor CsI(Tl) deposited onto the photodiodes. The physics of phosphors was described with reference to Fig. 2.4a. The photodiodes serve as the dels of the detector, detect the light emitted by the phosphor, and create an electrical charge signal that is stored on each del.

Because it can be manufactured to have a needle-like or columnar crystal structure, CsI can provide a better compromise between quantum efficiency and spatial resolution than is possible with the granular phosphors used in screen-film imaging. This is illustrated in Fig. 2.9b. In a conventional phosphor, the light quanta produced on X-ray absorption readily move laterally, leading to increased width of the line-spread function. The CsI crystals act as fiber optics or “light pipes” to reduce lateral spread. This allows the detector to be made thicker without as much resolution loss as would occur in conventional phosphors.

The arrangement of the individual dels with a photodiode and TFT switch is shown in the inset to Fig. 2.9a. Control lines for each row of the array are energized one at a time and activate all the switches in that row. A readout line for each column transfers the signal from the del at the activated row to an amplifier and digitizer. When a given row is activated, the signals from all of the dels on that row are collected along the readout line for all columns simultaneously.

In the system of this type, produced by General Electric Medical Systems (Milwaukee WI) (Fig. 2.10),

![Fig. 2.9. Flat panel detector with CsI(Tl) absorber. (a) Detector with photodiode array. TFT readout element is shown in inset. (b) Structure of CsI:Tl needle phosphor (Reprinted from Enhanced a-Si/CsI-based flat-panel X-ray detector for mammography, by Jeffrey Shaw, Douglas Albagli, Ching-Yeu Wei, and Paul R. Granfors; Medical Imaging 2004: Physics of Medical Imaging. Proc. SPIE 5368, 370 (2004). With permission from SPIE)](image1)

![Fig. 2.10. Photo of flat-panel detector. (Courtesy, GE Global Research Center)](image2)
the del pitch is 100 μm, the field size is 24 × 30 cm and the digitization is carried out at 14 bits (VEDANTHAM 2000). A comparison between the performance of the original detector system and one with an improved scintillating phosphor and reduced noise characteristics (SHEW 2004) was published by GHETTI (2008).

For flat-fielding correction, an offset value and a gain is measured for each del in the detector. Therefore, the number of such constants is equal to twice the number of dels in the detector, about 7.2 million values. It is typical to remeasure offset values between images; however, the gain matrix generally need only be measured occasionally.

### 2.7.2 Phosphor-CCD System

In this detector, an X-ray absorbing CsI(Tl) phosphor is deposited on a fiber-optic coupling plate, which conducts light from the phosphor to several rectangular charge-coupled device (CCD) arrays, arranged end to end. The fibers transmit the optical image from the phosphor to the CCD with minimal loss of spatial resolution. The CCD is an electronic chip containing rows and columns of light-sensitive elements. Light is converted in the CCD to electronic charge. The charge produced on each element in response to light exposure can be transferred down the columns of each CCD and read out by a single amplifier and analog-to-digital converter.

In the commercial implementation of this type of detector, the detector is rectangular with approximate dimensions of 1 × 24 cm. The X-ray beam is collimated into a narrow slot to match this format. To acquire the image, the X-ray beam and detector are scanned in synchrony across the breast (Fig. 2.11). Charge created in the CCD is transferred down the columns from row to row at the same rate, but in the opposite direction to the physical motion of the detector across the breast so that bundles of charge are integrated, collected, and read out corresponding to the X-ray transmission incident on the detector for each X-ray path through the breast. This is referred to as time-delay integration (TDI).

Scanning systems usually require longer total image acquisition time than full-field detectors. The slot collimators only allow use of a small portion of the total emission from the X-ray tube so that the overall heat burden for the tube for a scan is generally considerably higher than for full-field collimation. Because only part of the breast is irradiated at one time in scanning systems, the scatter-to-primary ratio is reduced. Collimation occurs before the breast so that transmitted X-rays are not lost. Normally, an antiscatter grid is not required with scanning systems while grids are used with full-field detectors. This provides a significant dose advantage for the former.

A slot-beam CCD-based scanning digital mammography system was originally marketed by Fischer Imaging Inc (Denver CO). It employs dels of 54 μm. Over a limited portion of the detector, data can be read out at 27 μm intervals to provide a high-resolution mode. Digitization is performed at 12 bits.

### 2.7.3 Photostimulable Phosphor (PSP) System

PSP systems, which are often referred to by their trade name, “computed radiography” or “CR,” have been widely used for many years in general radiographic applications. More recently, they were introduced for use in digital mammography. The operation of the detector in these systems is based on the principle of photostimulable luminescence, illustrated in Fig. 2.4b. Energy from X-rays is absorbed in a screen composed of a phosphor material containing a high prevalence of electron trapping sites. The absorbed energy causes electrons in the phosphor crystal to be temporarily freed from the crystal matrix and then captured in “traps” within the crystal lattice where
they can be stored with reasonable stability for times ranging from seconds to hours. The number of filled traps in a particular location is proportional to the amount of X-ray energy absorbed in that location of the screen.

This analog image is then read by placing the screen in a reading device where it is scanned with a red laser beam. This causes the electrons to be freed from the traps and to return to their original state in the crystal lattice. In doing so, they may pass between energy levels in the crystal structure. These energy levels are defined by small amounts of specific elements deliberately incorporated into the crystal. The choice of these materials thereby determines the color of the light emitted (related to the difference in energy between the levels) as the electron makes its transition. A typical strategy is to design the crystal to emit blue light, so that this can be measured with an appropriate optical filter placed in a light-collecting system incorporating a sensitive photomultiplier tube (Fig. 2.12a), without interference from the red laser light. The amount of blue light measured is proportional to the energy of X-rays absorbed by the phosphor.

The phosphor plate is continuous and is not physically divided into dels. The laser beam is scanned across the plate along one dimension as the plate moves through the reader in the orthogonal direction and the location of the beam on the surface of the plate at each point in time is used to define the x-y coordinates of the image. The spatial sampling is determined by the size of the laser spot (aperture, d) and the distance between sample measurements (pitch, p).

A photostimulable phosphor system for digital mammography was originally introduced commercially by Fuji Film. Several other PSP systems are now available\(^1\) and these are listed in Table 2.2. The dels are of a nominal size of 50 µm.

As discussed in connection with (2.2), (2.4) and (2.5), the gain, affects both the sensitivity and noise of the imaging system. It is important that the light produced from the phosphor is collected efficiently. If an inadequate amount of light is measured from each interacting X-ray, then the image will contain additional noise above and beyond the quantum noise, causing the SNR and DQE to be reduced.

To increase sensitivity and improve SNR, some photostimulable phosphor system manufacturers have refined their plate technology to reduce laser scattering and increased the efficiency of light collection by reading from both the top and bottom surfaces of the phosphor plate (Fig. 2.12b).

Unlike the other systems, this system employs removable cassettes, which can be used in the bucky tray of a standard mammography unit. While there are capital cost savings to this approach, it does require that phosphor plates be manually transported to the reader for processing. Because there are multiple detector plates, flat-field correction is normally not performed for the plates, but only for the plate reader. In principle, correction for nonuniformity of the individual plates could be done, but this would require precise registration within the reader and would be time-consuming.

---

\(^1\)Note that due to national approval procedures, some digital mammography systems are currently commercially available in certain countries but not in others.
### Table 2.2. Current digital mammography systems

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>Model</th>
<th>Del size (µm)</th>
<th>Detector dimensions (cm × cm)</th>
<th>Image matrix size (cm × cm)</th>
<th>Bit depth</th>
<th>Technology</th>
<th>Grid</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Flat panel detectors</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>GE</td>
<td>Senographe 2000 D</td>
<td>100</td>
<td>19 × 23</td>
<td>1,914 × 2,294</td>
<td>14</td>
<td>CsI on a-Si</td>
<td>Y</td>
</tr>
<tr>
<td>GE</td>
<td>Senographe DS</td>
<td>100</td>
<td>19 × 23</td>
<td>1,914 × 2,294</td>
<td>14</td>
<td>CsI on a-Si</td>
<td>Y</td>
</tr>
<tr>
<td>GE</td>
<td>Senographe essential</td>
<td>100</td>
<td>24 × 31</td>
<td>2,394 × 3,062</td>
<td>14</td>
<td>CsI on a-Si</td>
<td>Y</td>
</tr>
<tr>
<td>Lorad/Hologic</td>
<td>Selenia</td>
<td>70</td>
<td>24 × 29</td>
<td>3,328 × 4,096</td>
<td>14</td>
<td>α-Se</td>
<td>Y</td>
</tr>
<tr>
<td>Siemens</td>
<td>Mammomantation</td>
<td>70</td>
<td>24 × 29</td>
<td>3,328 × 4,084</td>
<td>14</td>
<td>α-Se</td>
<td>Y</td>
</tr>
<tr>
<td>Siemens</td>
<td>Inspiration</td>
<td>85</td>
<td>24 × 30</td>
<td>2,800 × 3,518</td>
<td>13</td>
<td>α-Se</td>
<td>Y</td>
</tr>
<tr>
<td>Planned Oy</td>
<td>Nuance</td>
<td>85</td>
<td>17 × 24</td>
<td>2,016 × 2,816</td>
<td>13</td>
<td>α-Se</td>
<td>Y</td>
</tr>
<tr>
<td>IMS</td>
<td>Giotto</td>
<td>85</td>
<td>24 × 30</td>
<td>2,816 × 3,584</td>
<td>13</td>
<td>α-Se</td>
<td>Y</td>
</tr>
<tr>
<td>Fujifilm</td>
<td>AMULET</td>
<td>50</td>
<td>24 × 18</td>
<td>3,540 × 4,740</td>
<td>14</td>
<td>α-Se</td>
<td>Y</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>with DOStech</td>
<td></td>
</tr>
<tr>
<td><strong>Scanning systems</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sectra</td>
<td>MDM L30</td>
<td>50</td>
<td>24 × 26</td>
<td>4,915 × 5,355</td>
<td>16</td>
<td>Si quantum</td>
<td>N</td>
</tr>
<tr>
<td>XCounter</td>
<td></td>
<td>50</td>
<td>24 × 30</td>
<td>4,800 × 6,000</td>
<td>16</td>
<td>Pressurized gas</td>
<td>N</td>
</tr>
<tr>
<td><strong>Photostimulable phosphor (PSP) systems</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fuji</td>
<td>Prefect</td>
<td>50</td>
<td>18 × 24</td>
<td>3,540 × 4,740</td>
<td>12</td>
<td>BaF(BrI):Eu</td>
<td>Y</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Carestream</td>
<td>DirectView CR950</td>
<td>50</td>
<td>18 × 23</td>
<td>3,584 × 4,784</td>
<td>12</td>
<td>BaFBr:Eu</td>
<td>Y</td>
</tr>
<tr>
<td>Agfa</td>
<td>CR 85 /35X</td>
<td>50</td>
<td>18 × 24</td>
<td>3,560 × 4,640</td>
<td>12</td>
<td>BaSrFBr1:Eu</td>
<td>Y</td>
</tr>
<tr>
<td>Konica</td>
<td>Pureview</td>
<td>43.8</td>
<td>35 × 43</td>
<td>~8,000 × 9,800</td>
<td>12</td>
<td>BaF:Eu</td>
<td>Y</td>
</tr>
<tr>
<td>Konica*</td>
<td>Regius 190</td>
<td>43.8</td>
<td>18 × 24</td>
<td>~4,300 × 5,800</td>
<td>12</td>
<td>BaF:Eu</td>
<td>Y</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Philips^</td>
<td>Cosima X Eleva</td>
<td>50</td>
<td>18 × 24</td>
<td>3,540 × 4,740</td>
<td>12</td>
<td>BaF(BrI):Eu</td>
<td>Y</td>
</tr>
</tbody>
</table>

*The Konica Regius 190 can use any of three possible plate types. Types RP-6M and RP-7M are based on BaF:Eu whereas type CP-1M uses a CsBr needle phosphor.

^The Philips CR unit uses the same plates as the Fuji CR unit.
2.7.4 Selenium Flat Panel

In this type of detector, the X-ray absorber is a thin layer (100–200 μm) of amorphous selenium. When X-rays interact with the selenium and produce energetic photoelectrons, these lose their kinetic energy through multiple interactions with electrons in the outer orbitals of selenium atoms. The process causes some of these electrons to be liberated and the freed electron and the corresponding “hole” created by its departure, i.e., the electron–hole pair, form the signal. An electric field applied between electrodes deposited on the upper and lower surfaces of the selenium as in Fig. 2.13a sweeps the charges toward the electrodes. One of the electrodes is continuous while the opposing one is formed as a large matrix of dels on a glass plate (YORKER 2002). The dels act as capacitors to store the charge. At the corner of each del is a TFT switch. Readout of charge from the dels is accomplished in the same manner as for the phosphor flat plate detector (Fig. 2.9), with control lines sequentially activating the TFTs for dels along individual rows. The signals from all activated dels are then simultaneously transmitted along readout lines adjacent to the columns of the matrix to be amplified and digitized. A detector of this type is produced by Hologic (Danbury CT). Dels are 70 μm, with 14-bit digitization. A selenium flat-panel detector system is also being produced by Anrad (St Laurent Quebec, Canada) with 85 μm dels (Fig. 2.13b) and this detector is currently used on the Giotto, Planmed, and Siemens systems. Some of its performance characteristics have been described by BISSONNETTE (2005).

Recently, an amorphous selenium detector that incorporates a different readout design (Fig. 2.14) has been introduced by Fuji. In this detector, there are two separate layers of selenium. The upper layer absorbs X-rays and produces electron–hole pairs similar to the operation of other selenium direct-conversion detectors. This charge is stored on the capacitance of each del. The lower selenium layer acts as an optically controlled switch that transfers the stored charge to a set of readout lines. This allows a del size of 50 μm to be achieved while avoiding the need for TFT switches, which would reduce the detector fill factor (Chap. 1).
in the conventional design and so reduce the geometric efficiency of the detector.

2.7.5 X-Ray Quantum Counting Systems

The detector systems described above operate by absorbing the energy from X-rays interacting with each del in the detector and accumulating the electronic signal produced by all the X-rays received during that measurement. This signal is then digitized to create the information corresponding to a pixel of the image. One aspect of these types of detectors is that higher energy X-ray quanta produce more signal in the detector than those of lower energy and this tends to weight the image signal to higher energy quanta. These carry comparatively lower image contrast than lower energy quanta.

Alternatively, the detector can be designed so that each del produces an electronic pulse every time an X-ray quantum interacts with it. The pulses are then counted to create the signal for that pixel. Pulse counting has several desirable features. Each interacting X-ray registers exactly one count regardless of its energy, so that the secondary noise sources discussed earlier associated with fluctuation in gain are eliminated. In addition, the equal weighting shifts the emphasis in the image signal away from the higher energies. Counting systems do not require the traditional analog-to-digital converter; however, it is important that the counting electronics is properly designed to handle the high rate of incident quanta, which can exceed $10^6$ per second.

Currently, two quantum-counting systems have been introduced. Both use a set of multiple linear detectors, which are scanned across the image field beyond the breast during image acquisition in synchrony with an appropriate set of collimator blades located on the X-ray entrance side of the breast. A precise mechanical scanning system is require in order to avoid image artifacts. The detector in the SECTRA system (Stockholm, Sweden) absorbs the X-rays in crystalline silicon (Fig. 2.15a). The electron–hole pairs produced from each interacting X-ray are collected in an electric field and shaped into a pulse, which is counted (ASLUND 2007). The XCounter (Stockholm, Sweden) employs a pressurized gas as the X-ray absorber and pulses of ions created in the gas form the signal as illustrated in Fig. 2.15b (THUNBERG 2002).

2.8 Spatial Resolution

Sample MTFs of several commercial digital mammography systems are compared with the MTF of a modern screen-film mammography system in Fig. 2.16. As will be discussed here, factors additional to the del aperture, d, also affect the MTF of detectors used for digital mammography. In some cases, these cause the ordering of the curves not to correlate with the del sizes.

The technical factors controlling spatial resolution differ among the types of detectors in use. In all phosphors, light is emitted from a small region near the point of X-ray interaction in the phosphor and tends to spread isotropically. In CsI(Tl) systems, where phosphor crystals are formed as columns (Fig. 2.9b), the crystals tend to guide the light down their length by total internal reflection and this considerably reduces spreading and allows the detector layer to be made thicker to obtain increased $\eta$. Even so, there is more spreading of light and a decrease in spatial resolution as the detector thickness is increased.

In photostimulable phosphor systems, the spread of light from its point of emission does not affect spatial resolution. The spatial localization is determined initially by the size of the laser spot that is used to read out the signal. As the laser light travels through the phosphor, it causes electrons trapped in the phosphor during the X-ray exposure to be freed from the traps. Some of these then produce light. The emitted light is collected by an optical system and a photodetector (Fig. 2.17a).

Sampling of the signal and the possibility of aliasing (discussed in Chap. 1) is controlled by the distance that the scanning laser spot moves relative to
the phosphor plate between light measurements. The laser light scatters in the phosphor so that traps are emptied, not only along the path on which the laser beam was directed, but also in adjacent regions of the phosphor corresponding to other areas of the image (Fig. 2.17b). This causes smearing of the signal and a loss of spatial resolution. The thicker the phosphor and the more scattering that takes place the greater the loss of resolution. Therefore, with these systems, there is a tradeoff between $\eta$ and spatial resolution.

Lateral spreading of signal can be reduced in direct conversion detectors when compared with phosphors because the charge signal is quickly swept toward the collection electrode by the electric field before the charge has much opportunity to spread (Zhao 1997). This offers the possibility of excellent spatial resolution with a detector that is thick enough to obtain a high value of $\eta$.

A more comprehensive and probably more relevant measure of imaging performance is obtained from the graph of DQE (see Chap. 1) vs. spatial frequency. Recall that DQE describes the efficiency of
the system in transferring the signal-to-noise ratio in the X-ray beam transmitted by the breast to the recorded image. One of the performance advantages of digital over screen-film mammography is the improved DQE. The improvement can be due to better signal transfer or reduced noise and this varies according to system design.

In Fig. 2.18, the DQEs of several digital mammography systems are compared with that of a modern screen-film detector. As can be seen, the DQEs for the digital systems are higher than for the screen-film detector at low to mid spatial frequencies. Because of the intrinsically higher spatial resolution of the screen-film receptor its DQE persists to higher spatial frequencies; however, at very low values. In a complete analog, mammography system the overall DQE would be brought down by the effect of the X-ray focal spot size.

More importantly, as illustrated in Fig. 2.19, the DQE given for the film system occurs only at the optimum

---

**Fig. 2.17.** Resolution loss in a photostimulable phosphor. (a) In the ideal system all emitted light arises from the point of X-ray interaction. (b) Scattering of readout laser light within the phosphor material causes blur.

---

**Fig. 2.18.** Spatial-frequency dependent DQE of screen-film mammography and some digital mammography systems (Data on GE systems from Ghetti et al. (2008), Fuji and Anrad systems from Rivetti et al. (2006), Selenia from Laz- zaria et al. (2007), Sectra from Monnin et al. (2007) and screen-film from Bunch (1999).

---

**Fig. 2.19.** Dependence of DQE on exposure to detector for a screen-film system (Bunch, 1999), a flat panel detector and a CR system (Data from Monnin, 2007).
exposure for the system. Ideally, the DQE of an imaging system should be independent of exposure to the detector, so that the value of SNR\(^2\) should be directly proportional to exposure. For films systems, DQE is reduced considerably by the effects of film granularity (at low exposures) and nonlinearity of the response curve of the film (at both low and high exposures). For digital mammography detectors, there tends to be less dependence of DQE on exposure although there is generally some effect. For systems where there is nonuniformity of sensitivity across the image field (e.g., CR systems where flat-field correction is not performed), DQE falls slightly toward the high end of the exposure range. For systems where there is a significant signal-independent level of electronic noise, DQE will fall at low exposures where the effect of this noise on the SNR is most important.

Several authors have performed comparison measurements of various performance indices such as MTF and DQE of commercial digital mammography systems, e.g., (Lazzaria 2007; Rivetti 2006; Monnin 2007).

2.9 Toward Smaller Dels

One feature of digital mammography that initially, at least, caused concern among radiologists, is that because of the del sizes used in the systems, the limiting spatial resolution would not be as high as when a high-quality screen-film receptor were used. For example, imaging a Pb line-pair test pattern with an excellent screen-film system might allow as many as 22 line-pairs/mm to be resolved. The basic design of digital systems suggests that maximum spatial resolution would be on the order of only 5 lp/mm for a 100 \(\mu m\) del and 10 lp/mm for a 50 \(\mu m\) del. Clearly, the expected advantages of digital mammography must come from something other than limiting spatial resolution, i.e., better contrast, image processing, quantitative information, improved archiving and retrieval, etc.

Nevertheless, one can ask, why not design a digital mammography system with smaller dels and obtain improved limiting spatial resolution? There are many different answers to this question. Perhaps, the first comes from asking another question: Is it necessary or desirable to have smaller dels? While the DMIST study illustrated a sensitivity advantage of digital mammography over film, primarily in women with dense breasts, many radiologists feel that in some cases, microcalcifications are not as well visualized with digital mammography. Magnification digital mammography, a technique accomplished by moving the breast closer to the X-ray source, causes the effective size of the del to be reduced with respect to the size of anatomical features and has been found to improve lesion conspicuity. This may also be the case with fine fibrils radiating from tumor masses. This suggests that with all other factors remaining equal, there may be an advantage to reducing the size of the del. Of course, in reality, the usual dilemma associated with any engineering problem then occurs, i.e., when one aspect of imaging is improved, another is generally degraded.

In flat panel systems, several characteristics of the performance of the detector change when the del size is reduced: (1) generally the fill factor of the del is reduced because an increased proportion of the del area must be used to accommodate switches and readout and control lines, (2) the charge storage capacity under the del is reduced, and related to this (3) the readout noise becomes larger when compared with the signal produced. In addition, there are nondetector factors that come into play and these may argue against reduction of the del size. For a given X-ray exposure to the breast, fewer quanta will fall on and be captured by a smaller del, causing the quantum SNR per del to be reduced. The amount of data produced per image increases inversely as the square of the linear dimension of the del (reduction of del dimension by a factor of 2 causes images to contain four times as many pixels). This increased amount of data must be read out at an acceptable rate and processed by the subsequent components of the imaging system. Finally, the production yield of detectors with an acceptable number of properly operating dels falls as the del size is reduced and this increases the cost of the detector. Nevertheless, many of these factors are technology-related rather than being fundamental and are therefore amenable to being overcome. The system described in Fig. 2.14 is an example of one approach to a solution.

2.10 Automatic Exposure Control

Digital image acquisition provides opportunities for major improvement in automatic optimization of image acquisition. For example, it is no longer necessary to have a separate AEC sensor as part of the system because the digital detector can serve as a multi-element sensor. This is currently not possible
with the photostimulable phosphor systems because they are used with a conventional film mammography machine and rely on the AEC that is part of the unit.

The optimal way to perform AEC in digital mammography is still a subject under study; however, several approaches have already been put forward. One method involves acquiring a test image of the breast at very low dose and using data gathered from this image to determine the exposure parameters that will be used for the main exposure. In the implementation used by Siemens, for example, the dose is kept very low for this test pulse by binning together data from many dels (e.g., 128 × 128) to create “super-pixels.” After the exposure, either a manually preselected region of interest (ROI) from the detector data is used or else an algorithm is employed to segment the entire area of the breast from the image and this is used as the ROI. A goal is defined to specify the requirements for a high-quality image of the breast. For example, this could be that the mean pixel value over the ROI is at a specified level or that no pixel value in the ROI is less than some specified value, etc. From the statistics of the test image, the exposure requirements for the actual image can be inferred. These settings would be automatically selected immediately after the test exposure and the main exposure would then occur.

To use the detector in this way, it is necessary that the detector can be read out quickly to determine what the optimum exposure factors should be. In addition, the detector response must be stable at the low exposures used for the test shot and the detector response at different X-ray fluence levels must be sufficiently predictable to allow scaling of the settings from those used during the test shot to those required for the main exposure. This may impose difficulty for some of the current detectors and creative ways of accomplishing this will have to be found.

One approach to automatic optimization of exposure is to compute one or more appropriate statistics that reflect the attenuation of the breast from a brief, low-dose test exposure obtained immediately prior to the actual image acquisition. One possible statistic might be the minimum signal. This would arise from the most attenuating region of the breast. The algorithm could then set the parameters for the actual exposure so that the image signal in this region would exceed some preset value. Because, in a digital image signal is less important than SDNR, it might be more useful to have the algorithm compute the SDNR from the test exposure and then ensure that some minimum acceptable value of SDNR is exceeded in every part of the image. Additional data on the compression thickness and force can also be used to refine the algorithm. There is likely considerable opportunity to improve imaging performance in mammography through design of “smarter” AEC systems.

References