fundamentals of Digital Mammography

Review

Physics,

Technology

& Practical

Considerations

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Contents

Introduction

While screen-film has been the standard detector used in conventional mammography systems, new developments in detector and computer technologies are altering the landscape for mammography imaging. In particular, digital mammography detectors using selenium promise to revolutionize the field. New flat-panel x-ray detectors based on amorphous selenium technology offer extremely high quantum efficiency and high resolution, resulting in mammograms with potentially lower dose, better image quality, and improved diagnostic performance.

This review describes the technical basis for current and future advances in digital mammographic detector technology. These include the following:

- Lower dose
- Improved image quality
- Computer-aided diagnosis
- Soft-copy review and digital archiving
- Tomosynthesis and other three-dimensional visualization techniques

Digital Detector Technologies

Digital detectors for mammography can be categorized as indirect or direct. Indirect digital detectors utilize an indirect method of imaging X-rays, similar to screen-film. A scintillator absorbs the X-rays and generates a light scintillation that is then detected by an array of photodiodes. Indirect systems suffer from resolution degradation caused by light spread in the scintillator and poor quantum efficiency with thin scintillators.

Direct digital detectors utilize a direct-conversion method of imaging, wherein the X-rays are absorbed and the electrical signals are created in one step. Systems using amorphous selenium represent a direct technology for digital mammography. Selenium is an ideal material for a mammography detector because it has high x-ray absorption efficiency, extremely high intrinsic resolution, low noise, and a well-established manufacturing process.

Field of View

The field of view for full-field digital mammography (FFDM) systems is very important. In order to image most of the adult female population, the imaging field of view must be similar to the size of the largest screen-film cassette, 24 x 30 cm.

Pixel Size

Spatial resolution depends on pixel size and the conversion method of the detector. Even with pixel sizes below 100 µm, the spatial resolution of indirect detectors is limited primarily by light blurring in the scintillator. Direct systems do not have such limitations. Using a 70-µm pixel size, a selenium detector can meet the high-resolution requirements of mammography while offering reasonable costs in terms of manufacturing and hospital information resources. An advantage of selenium detectors is that the pixel size can be reduced to keep pace with manufacturing, network, and display technology advances; indirect detectors will not benefit from decreased pixel size due to scintillator limitations.

Detector Performance

The detector performance measures of modulation transfer function (MTF) and detective quantum efficiency (DQE) are introduced. MTF is a measure of resolution, and DQE is a measure of dose efficiency and signal-to-noise ratio. The MTF and DQE of direct detectors are superior to those of screen-film and indirect digital detectors. With superior DQE at all spatial frequencies, selenium detectors offer both improved image quality and lower patient dose.

Digital images offer a variety of new and improved applications. Contrast enhancement and the wide dynamic range of digital detectors will improve visibility of mammographic features. The digital image will provide archiving and retrieval advantages over film, and will facilitate the use of computer-aided diagnosis.

Future Applications

Future applications such as stereo mammography, tomosynthesis, and other three-dimensional imaging modalities are under investigation. Such advances in technology will provide improved diagnostic information and reduced image confusion from overlapping structures. These three-dimensional imaging tasks will benefit from the high quantum efficiency that direct detectors offer. Other promising applications include full-field and high-resolution stereotactic breast biopsies and diagnostic imaging. However, it is essential that full-field digital systems perform well for the primary task of breast screening.

In conclusion, selenium-based direct digital detectors provide ideal mammographic performance. The superior performance of selenium FFDM detectors should result in improved image quality, lower dose, and improved diagnostic accuracy.

Screen-Film Mammography

Conventional film systems use intensifying screens to capture X-rays and reduce radiation dose. These screens are often constructed of rare earth phosphors such as gadolinium oxysulfide (Gd_2O_2S). X-rays that pass through the tissue are collected by phosphor screens and converted to light. Film in close proximity to the screen captures the light photons, and the image is obtained by exposing the film (Figure 1).

While thicker screens capture more X-rays, they also create more light scatter and blur the image. Therefore, it is impossible to offer a screen-film system simultaneously offering the highest possible resolution and lowest possible radiation dose. This trade-off between radiation dose and image quality must be optimized for the specific clinical application.

Figure 2 shows the performance trade-off between sensitivity and resolution inherent in the design of screen-film systems. When an X-ray is absorbed, the resultant light scintillation creates a number of light photons that spread and illuminate the film in a distribution cloud. As the screen is made thicker, the cloud of light on the film will increase in size. This reduces the resolution of the system, but increases sensitivity since more incoming X-rays are absorbed. The opposite occurs when screen size is decreased.

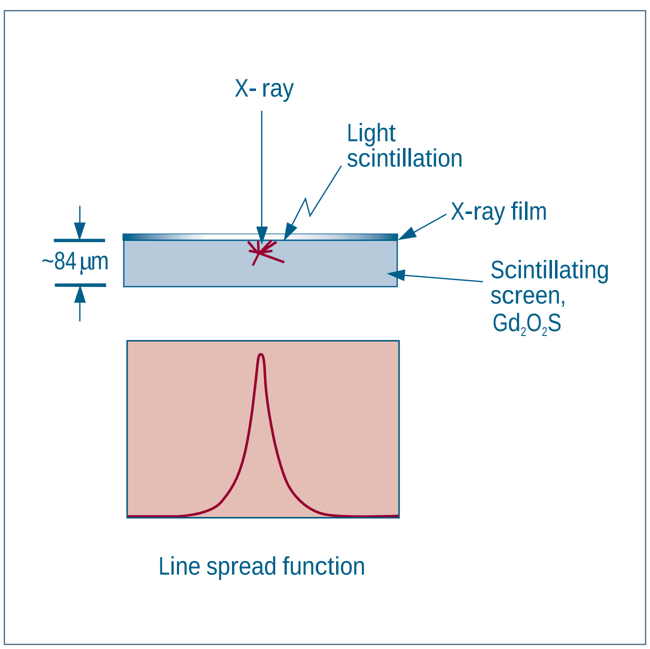

Figure 1. Screen-film systems use a scintillating screen to absorb X-rays and generate light photons, which are captured on the film.

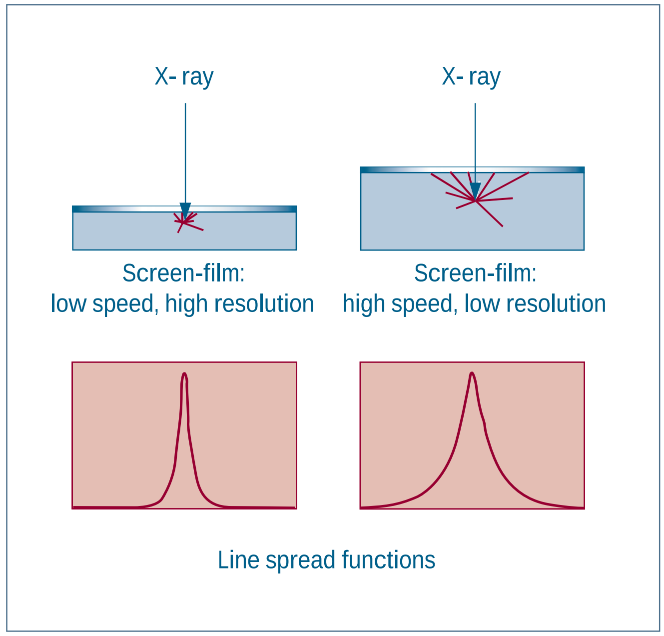

Figure 2. As the thickness of the screen is increased, its speed increases at the expense of decreased spatial resolution.

In a screen-film mammography system, the film is placed at the entrance surface of the intensifying screen. Because X-rays are absorbed with a decaying exponential spatial distribution, more X-rays are absorbed near the entrance of the intensifying screen than at its exit. Placing film on the entrance side maximizes spatial resolution.

While film offers several advantages, there are significant disadvantages and limitations. Film does not have a linear sensitivity to photon flux, and there is a narrow range over which it can detect small differences in contrast. Because of film's limited latitude, tissue areas of high and low density are often sub-optimally imaged. Frequently, the image is poorly exposed because of film's stringent requirements for proper exposure, resulting in repeated exposures. Resulting images often suffer from granularity problems, reducing visibility of microcalcifications.

Film also requires processing time and storage space. Chemical processing adds time to the exam, and the resulting x-ray films require a large amount of storage space in patient health records. Film also must be physically transported to the physician for viewing.

Digital Mammography Detectors

Digital technology offers the potential for several advances in mammography. Because images are captured as digital signals, electronic transfer and storage of images is possible, eliminating physical storage and distribution required by film. Digital systems offer a large dynamic range of operation, improving visualization of all areas of the breast and increasing exposure latitude. Also, the digital format allows grayscale adjustment to optimize contrast for every imaging task. Soft-copy reading, computer-aided diagnosis and three-dimensional imaging offer other important opportunities for improvement in mammographic systems.

Digital Detector Technology

There are two methods of image capture used in digital mammography that represent different generations of technology: indirect and direct conversion.

Indirect Digital Detectors

Indirect digital mammography systems use indirect-conversion detectors, as shown in Figure 3. Such detectors use a two-step process for x-ray detection. The first step requires a scintillator, such as cesium iodide doped with thallium [CsI(Tl)], to capture the x-ray energy and convert it to light. An array of thin-film diodes (TFDs) converts light photons to electronic signals that are captured using thin-film transistors (TFTs). Some systems use charge-coupled devices (CCDs) as an alternative light collection array and readout method.

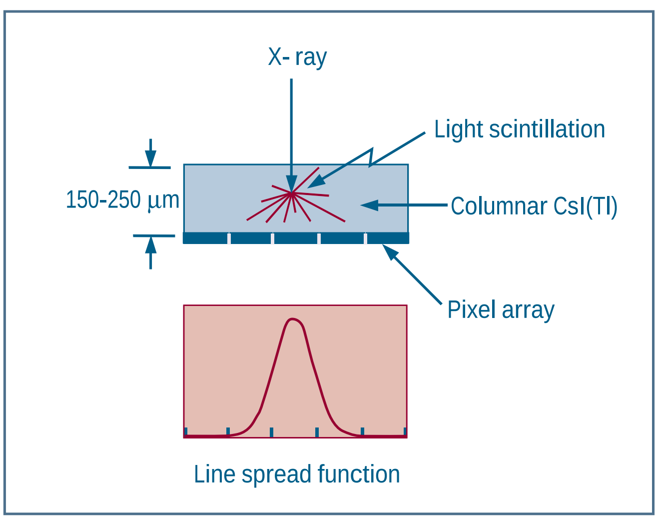

Figure 3. Indirect detectors utilize a scintillating layer to absorb X-rays and generate light photons, which are detected by a photodiode array.

Similar to screen-film, light scatter compromises image quality, and there is a performance trade-off between spatial resolution and radiation sensitivity. As the scintillator is made thicker, light spread increases, resulting in decreased resolution (Figure 4). Because of its columnar structure, CsI(Tl) does not create as much light scatter as other screens. However, the problem of compromising between resolution and sensitivity still exists.

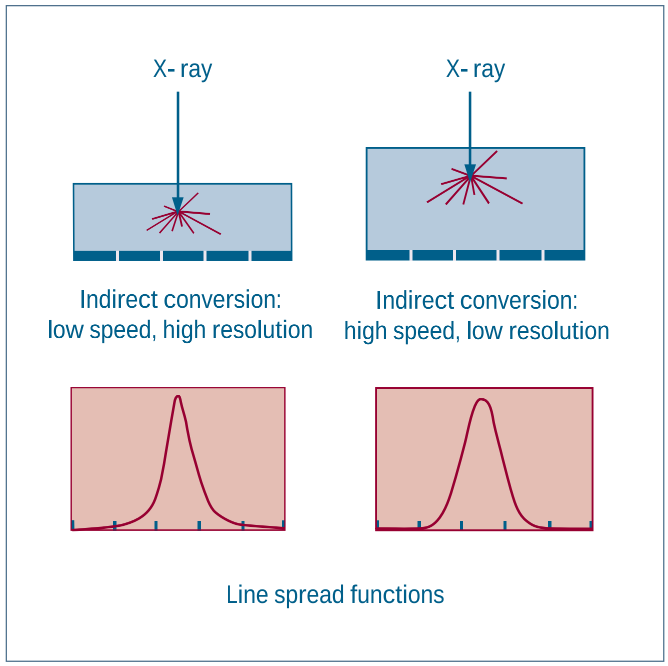

Figure 4. As scintillator thickness increases, resolution decreases in indirect detectors.

The placement of the scintillator is more problematic in indirect digital detectors than with screen-film systems. As with screen-film systems, more X-rays are absorbed near the entrance of the scintillation layer than the exit. While film is placed near the entrance side of the scintillator, a photodiode-transistor array is not transparent to X-rays, and the array must be placed on the exit surface of the scintillator. This causes more degradation in spatial resolution compared to screen-film.

The typical thickness of CsI(Tl) used in mammography detectors ranges from 150 to 250 µm. These indirect digital detectors exhibit light spreading similar to screen-film systems.

Direct Digital Detectors

Direct digital detectors represent a technological advance, eliminating problems associated with light scatter inherent in indirect systems (Figure 5). In direct systems, a photoconductor absorbs the X-rays and directly generates the electronic signal, without intermediate steps that degrade image quality (direct conversion). Under the influence of an external electric field, holes (or electrons, depending upon the polarity of the applied field) drift towards a pixel electrode and are collected on a pixel capacitor. Because the electrons and holes travel along electric field lines, there is no lateral movement of the charge. This results in an exceptionally narrow point spread response of about 1µm.

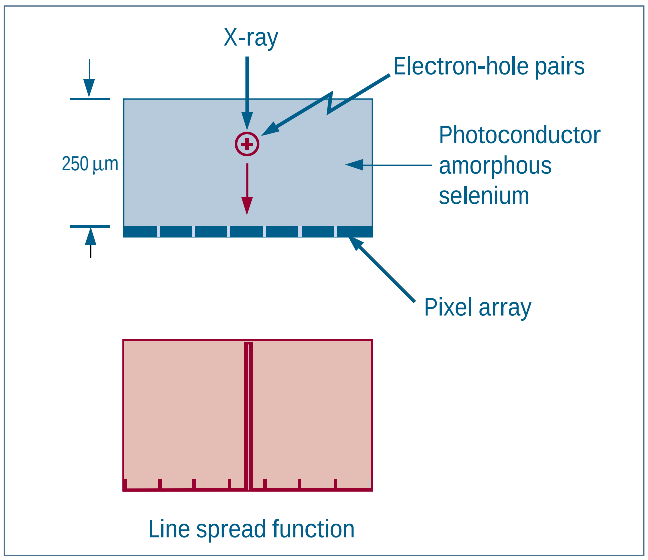

Figure 5. Direct detectors use a photoconductor to absorb the X-ray and directly generate the signal.

The photoconductor of choice used in direct systems is amorphous selenium (a-Se). Selenium has a long commercial history in xeroradiography, and its manufacturing processes are well known and optimized. Although the image quality of xeromammography systems was widely acknowledged, they suffered from reliability problems from mechanical wear of plates during toner deposition and crystallization of selenium during high-temperature erasure cycles. By depositing selenium on a flat-panel electronic imaging receptor, these problems have been eliminated.

In direct digital detectors, the response function maintains its sharpness even as the thickness of the photoconductor is increased, so there is no trade-off between radiation stopping power and spatial resolution (Figure 6). In practice, the photoconductor is made sufficiently thick in order to stop most of the incident X-rays. As the photoconductor is made very thick, parallax errors due to non-oblique incidence contribute some degradation to the intrinsic resolution.

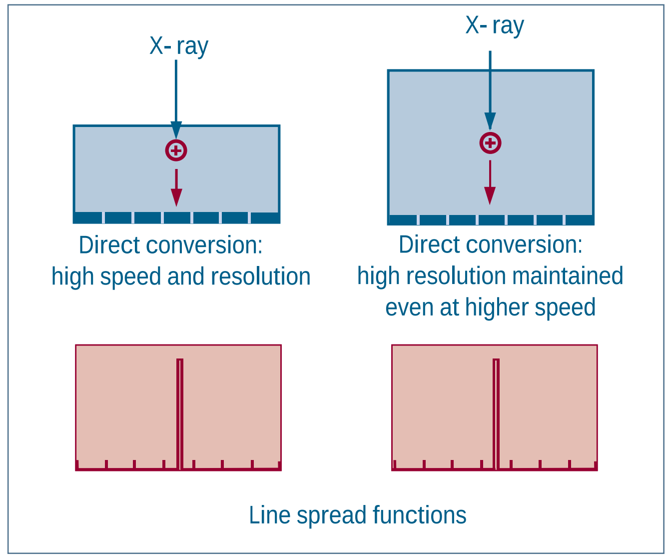

Figure 6. There is no trade-off between spatial resolution and sensitivity for direct digital systems.

However, this effect is small since most of the X-rays are absorbed near the entrance surface of the detector.

Using amorphous selenium as the photoconductor, a thickness of 250 µm is adequate to stop >95% of the X-rays in the mammographic energy range. Standard screens for film mammography only have about 50-70% quantum efficiency. Indirect digital detectors with CsI(Tl) exhibit about 50-80% quantum efficiency. Due to the location of its K-edge just below the diagnostic range, the selenium system achieves almost complete quantum efficiency (Figure 7).

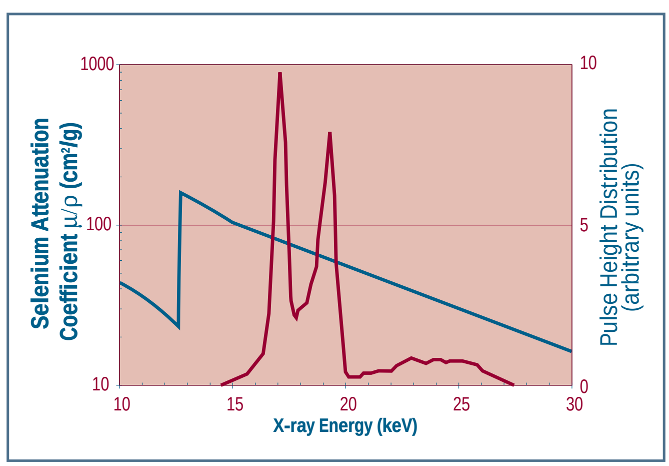

Figure 7. Absorption coefficient for selenium (in blue) shows the K-edge just below the diagnostic range for mammography. A typical Mo-Mo 28 kVp mammographic spectrum incident on the detector is shown in red.

Pixel Design for Digital Detectors

Digital detectors require an array of pixels that collect electronic signals. The signals on these pixels are transferred to a computer during a readout sequence. This is known as direct readout, a function of all digital systems, and should not be confused with direct-conversion digital detection.

TFT or TFD arrays are commonly used as the active electronic readout mechanism in both direct- and indirect-digital radiography systems. The arrays are typically deposited onto a glass substrate in multiple layers, beginning with readout electronics layer on the glass, followed by charge collector arrays. The composition of the top layer depends upon the type of detector. If the system uses indirect conversion, both x-ray elements and light-sensitive elements are deposited on the top layer. Direct detectors do not require conversion of X-rays to light, so light sensitive elements are not necessary for these systems.

Direct TFT arrays are considerably easier to fabricate than TFD arrays in indirect detectors. These direct selenium-based detectors do not require a photodiode construction. These systems also utilize the same manufacturing processes as large-area liquid crystal displays (LCDs), commonly used in computer monitors.

Charge-coupled devices (CCDs) are an alternative to TFT arrays in indirect-conversion systems. Basic CCD-based systems consist of a series of metal oxide semiconductor capacitors that are fabricated close together on the semiconductor surface. These systems use fiber optics to capture light emitted from scintillators or intensifying screens, but suffer from signal loss, added complexity of fiber optics, and the necessity to piece together separate CCD arrays.

Pixel Size Considerations

Digital detectors are composed of arrays of pixels. The smallest feature that can be resolved in any digital imaging system is a function of the pixel size - the smaller the pixel, the higher the resolution. In indirect digital detectors, as the pixel size is decreased, a limit is reached beyond which further reductions in pixel size do not significantly improve resolution. This resolution limit results from light scattering in the scintillator, which exhibits line spread widths of about 100 µm (Figure 8). Pixel sizes smaller than 100 µm do not offer significant improvements in resolution, and the performance of such indirect systems is barely adequate for digital mammography.

In a direct digital detector, spatial resolution is limited only by the size of the pixel (Figure 8). The size of the pixel

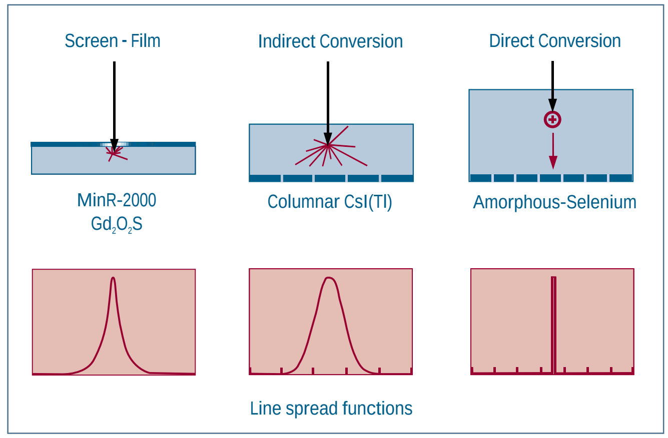

Figure 8. Line spread functions for screen-film and digital detectors. Screen-film and indirect detectors have broad line spread functions; direct detectors have narrower line spread functions.

in these detectors can be made arbitrarily small to extend the resolution performance to very high spatial frequencies. Since microcalcifications can be as small as 100 µm, pixels larger than 100 µm do not provide adequate morphological information on the smallest objects of interest in mammographic imaging.

In addition to requirements for diagnostic imaging, the optimal pixel size is determined by limitations of information and display systems and manufacturing processes. As the pixel size of any digital imaging system is made smaller, the amount of data contained in the image rapidly increases. This increases hospital information system costs in terms of data storage, network bandwidth, and display capabilities. Also, fabrication of large-area detectors with very small pixels is an expensive, low-yield process.

The pixel size in the DirectRay® selenium detector is 70 µm. This pixel size offers reasonable costs in terms of manufacturing and hospital information resources and meets the high-resolution requirements of mammographic imaging.

Field of View Requirements

The Mammography Quality Standards Act (MQSA) requires that mammography facilities use at least two screen-film cassette sizes: 18 x 24 cm and 24 x 30 cm. The rationale for this requirement is the large distribution in compressed breast sizes. A small cassette cannot accommodate large breasts in the field of view, and two overlapping exposures will be required to ensure adequate coverage of all the breast tissue. This is costly, time-consuming, and subjects the women to additional radiation dose.

Digital mammography presents additional field-of-view issues. Flat-panel x-ray detectors are considerably more expensive than screen-film cassettes, precluding the use of more than one receptor size. Aside from cost, digital detectors are heavy and fragile and do not lend themselves to rapid changeover like Buckys. Consequently, digital mammography manufacturers must decide on a detector size large enough to accommodate most women without requiring double exposures.

A digital detector 18 x 24 cm (the smaller film cassette size) is inadequate to image as many as 15-30% of US women, depending upon regional differences.* One solution to this problem would be to employ digital image stitching. In this scenario, breasts too large to image on the small detector in one exposure would undergo multiple exposures, and the computer would merge or "stitch" these into one simulated exposure of the entire breast. Disadvantages of this include additional setup and imaging time, and breast regions that receive repeated radiation exposure. More importantly, it would be virtually impossible to obtain a seamless image without significant distortion in the overlap region, due to patient motion or positioning errors.

Use of a larger detector for digital mammography is an important design issue and, in fact, a significant feature. Using a large digital detector to image a small breast presents no difficulties in positioning since the breast does not have to be centered over the photodetector to obtain the correct exposure. The automatic exposure control system looks at the entire image area and then selects the area under the densest portion of the breast as the area to be used for determining the correct exposure. (In addition, the long, linear response of the digital detector means it is not as sensitive as film to small variations in exposure.) The larger image area can then be cropped to reduce data transfer time, display requirements, and storage space. Most importantly, a digital detector with a field of view of approximately 24 x 30 cm can be used to image almost all mammography patients, unlike smaller detectors.

Digital Mammography Detectors in Development

Table 1 lists some full-field digital mammography detectors in development.[1-3]

	Indirect			Direct
	GE	LORAD†	Fischer Imaging†	LORAD (DirectRay)†
Conversion Method	Indirect	Indirect	Indirect	Direct
Scintillator	CsI(Tl)	CsI(Tl)	CsI(Tl)	amorphous selenium
Photoconductor	TFT	CCD	CCD	TFT
Pixel Size	100 µm	40 µm	24-48 µm	70 µm
Field of View	18 x 23 cm	19 x 25 cm	22 x 30 cm	25 x 29 cm (Approx)
Detector Area	414 cm^2	475 cm^2	660 cm^2	720 cm^2 (Approx)

Table 1. Digital Mammography Detectors †Not available for sale in the US at this time.

* *Based upon historical sales data for mammography film.*

Characterizing Detector Performance

There are several well-accepted parameters that measure the overall image quality of mammography and all other radiography systems. These are:

- Contrast resolution
- Signal-to-noise ratio
- Dose efficiency
- Spatial resolution

Modulation transfer function (MTF) and detective quantum efficiency (DQE) provide quantitative measurements of imaging performance. MTF measures image sharpness, while DQE is a measure of signal-to-noise ratio, contrast resolution, and dose efficiency. Imaging performance in radiography is best characterized by examining corresponding MTF and DQE curves; however, performance cannot be adequately described by one number at a single spatial frequency. These measurements are used to determine how well a system captures information over a range of spatial frequencies.

Modulation Transfer Function (MTF)

MTF is a measure of signal transfer over a range of spatial frequencies and quantifies image sharpness.

Indirect-conversion methods can scatter light over several pixels (as shown in Figure 8), further limiting the effective resolution of the system, more so than indicated by pixel size alone. Direct-conversion systems do not suffer from this limitation. As shown in Figure 9, the MTF of a selenium direct-conversion detector[3] is superior to those of screen-film[4] and indirect-conversion detectors.[1, 2, 5]

The intrinsic image sharpness of selenium direct-conversion mammography detectors is superior to those using indirect-conversion scintillators. While MTFs for indirect detectors fall dramatically at higher spatial frequencies, the MTF of a selenium detector remains high over a greater range of spatial frequencies. With selenium, there is no lateral movement of the charge through the photoconductor, and its MTF is independent of thickness. Consequently, selenium detectors are very efficient in capturing and converting X-rays to electrical signals.

Detective Quantum Efficiency (DQE)

Even with a high MTF at high spatial frequencies, small objects can get lost in the noise of the system. Increasing signal and decreasing noise in the system increases visibility of small structures. DQE measures signal-to-noise transfer through the system as a function of spatial frequency, and it is a good measure of dose efficiency. Several factors influence DQE, including the amount of x-ray absorption, amplitude or strength of the signal profile (measured by MTF), and noise.

While film has a high MTF at high spatial frequencies compared to the indirect detectors, the same is not true for DQE (Figure 10). This is one reason why the potential spatial resolution with film is not realized in practice. Film granularity noise limits its achievable DQE at high spatial frequencies. Although indirect systems have a DQE superior to that of screen-film, especially at the lower spatial frequencies, the DQE drops at high spatial frequencies. This is a consequence of scintillator-induced light blurring.

Selenium digital mammography detectors outperform both indirect digital and film in terms of DQE. With no signal spreading, the DQE (and MTF) are governed mainly by the inherent limits of the pixel size. Furthermore, only selenium direct systems can be made sufficiently thick to absorb all the incoming X-rays without sacrificing spatial resolution.

Direct systems offer the *potential* of similar image quality at lower doses or improved image quality at the same dose as screen-film. From a clinical point of view, there is the expectation that at equal doses, direct FFDM systems may provide superior diagnostic image quality.

Dynamic Range

Screen-film has a limited dynamic range, which prevents visualization with equal clarity of all breast tissue regions from the chest wall to the skin line. Figure 11 shows the need for improved dynamic range in breast imaging.

Digital detectors offer great improvements in dynamic range. For an ideal detector with no inherent noise, 3100 gray levels are discernable in a typical mammographic image.[6] A system that provides at least 12 bits of dynamic range will not degrade the underlying information. Consequently, digital mammograms exhibit uniform quality over a wide range of exposures.

X-ray Quantum Detection Efficiency

X-ray quantum detection efficiency measures the percentage of X-rays hitting the detector that get absorbed. Systems with higher quantum efficiency can produce higher quality images at lower doses. As mentioned earlier, selenium systems stop greater than 95% of the X-rays in the mammographic energy range. Standard screens for use in film mammography only stop about 50-70%, and CsI(Tl) used in indirect detectors stop about 50-80%.

An important issue with the DQE is its dependence on exposure. The DQE of screen-film systems changes rapidly as exposure, or film density, is varied and has a maximum value at a density of about 1.20. At higher densities, such as those used in mammography, the DQE is significantly lower than that shown in Figure 10.[5]

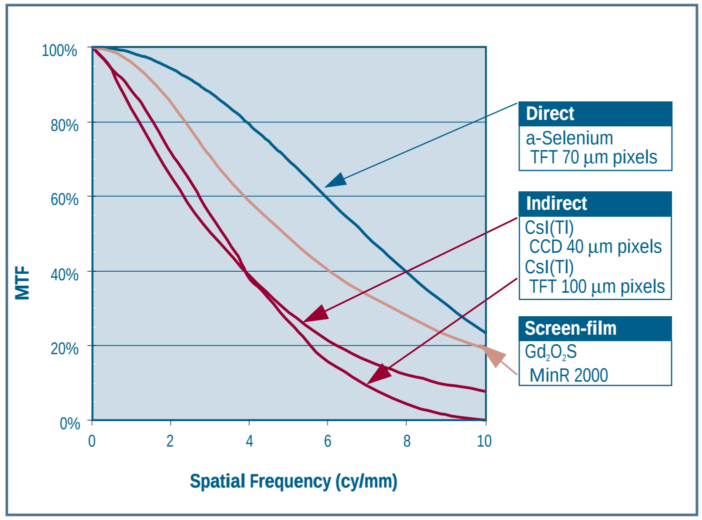

Figure 9. Modulation transfer functions for screen-film (Kodak MinR 2000), indirect (GE Senographe 2000D and LORAD DBI) and direct (DirectRay) digital mammography detectors.

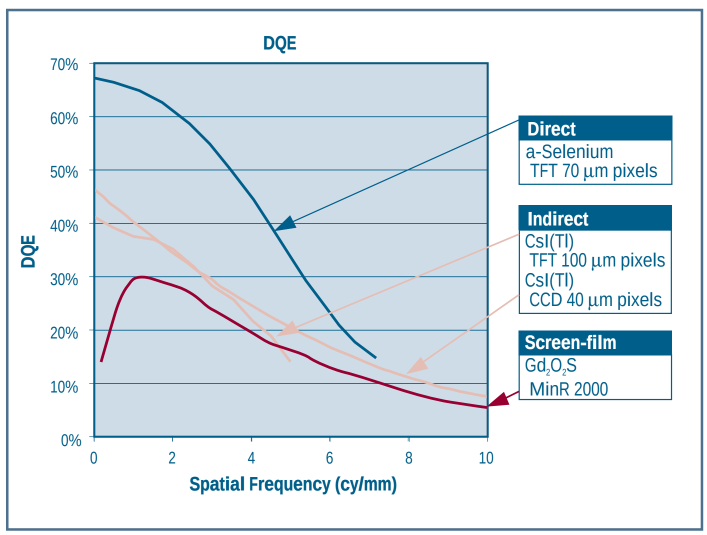

Figure 10. Detective quantum efficiency for screen-film (Kodak MinR 2000 at 10 mR), indirect (GE Senographe 2000D at 11 mR and LORAD DBI at 8 mR) and direct (DirectRay) digital mammography detectors.

ACR Phantom Imaging Performance

The ACR phantom is approximately equivalent in x-ray absorption to a 4.2-cm thick compressed breast consisting of 50% glandular and 50% adipose tissue. The phantom includes appropriate details that range from visible to invisible on a standard mammographic film image.

The visibility of phantom details has been evaluated for screen-film and for indirect and direct detectors.[7] The phantom was imaged at 28 kVp at a dose of ~3 mGy (300 mrad). Results are summarized in Table 2. These results are consistent with predicted performance based on the DQE: direct-conversion systems outperform both screen-film and indirect-conversion systems.

It is important to note that it is possible with the direct-conversion systems to see all of the objects in the ACR phantom, i.e., a total of 16 fibers, speck groups, and masses. This is not possible with screen-film or indirect-conversion systems.

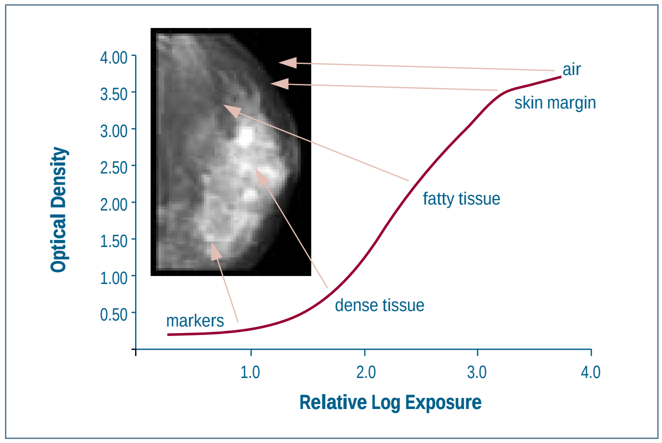

Figure 11. Characteristic curve of mammographic film illustrating how the display contrast (slope of the curve) is sub-optimal in lucent and dense regions of the breast. Adapted from Yaffe M. *Digital Mammography, RSNA Categorical Course in Breast Imaging*: 1999:229-238.

Detector	Technology	Fibers Visible	Specks Visible	Masses Visible	Total
Kodak MinR 2000	Screen-film	4	4	3	11
LORAD CCD	Indirect Conversion	5.5	4	5	14.5
LORAD a-Selenium	Direct Conversion	6	5	5	16

Table 2. Visibility of phantom details for film and digital mammography detectors.

System Design Considerations

There are a variety of system-level design considerations with full-field digital mammography equipment.

X-ray Source

While the spectrum of x-ray energies used in screen-film mammography has been highly optimized, digital mammography detectors offer performance improvements, especially in dynamic range. However, the issue of the optimum energy is still in the investigational phase.[8] Higher x-ray energies may permit lower dose or higher image quality with digital mammography, particularly for patients with dense breasts. Because direct selenium detectors have high intrinsic quantum efficiency, the use of higher energies is possible.

Selenium has superior stopping power compared to film screens and indirect CsI(Tl) scintillators (Figure 12). At typical energies used in mammography (mean energy ~ 20 keV), selenium's quantum efficiency is optimal. Selenium maintains its high quantum efficiency at higher energies as well. Because systems that reduce breast compression will require higher x-ray energies than used currently, the pronounced high-energy performance of selenium will become even more important.

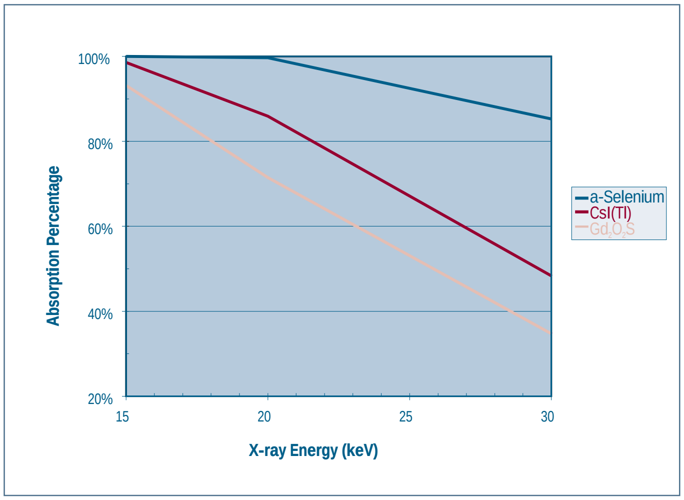

Figure 12. Percent absorption of incident X-rays for materials used in screen-film Gd_2O_2S (90 µm thick), direct conversion CsI(Tl) (100 µm thick), and indirect conversion selenium (250 µm thick)

Scatter Rejection Methods

Digital detectors made from area detectors are subject to the same deleterious effects of radiation scatter as conventional screen-film. The preferred method to reduce scatter is to employ the use of radiation anti-scatter grids interposed between the patient and the detector.

Two types of grid construction are illustrated in Figure 13. Standard linear grids are constructed of long thin strips of radio-opaque materials (called septa or laminae) separated by radiolucent spacers. These grids effectively reduce scattered photons, but also block some of the primary beam, mainly from absorption in the spacers. The high transmission cellular (HTC) grid absorbs scattered radiation in two directions, as opposed to the linear grid, which removes scattered radiation only in one direction. Because the interspace material is air, this grid absorbs less of the primary beam. As shown in Figure 14, the HTC grid has superior contrast improvement (scatter rejection) and Bucky factors (primary beam reduction) compared with conventional linear grids.[9]

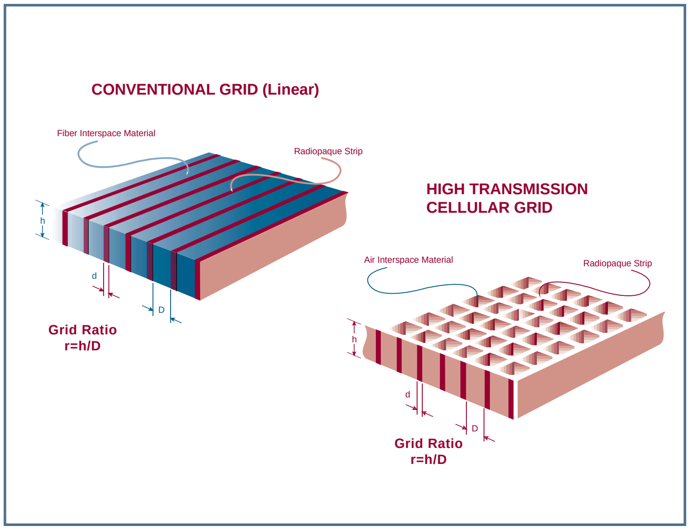

Figure 13. Construction of linear and high transmission cellular anti-scatter grids.

Automatic Exposure Controls

Breasts vary in composition and thickness, and there is a need for automatic exposure control (AEC) to optimize image quality and reduce radiation dose for each patient. Because of film's narrow exposure latitude, proper AEC operation is critical. Digital detectors have a far wider dynamic range and consequently are more tolerant of exposure variations. Therefore, digital mammography should have a reduced number of retakes compared to screen-film. Digital mammography systems also offer the possibility of more advanced AEC methods. It is possible to design these systems to acquire a rapid low-dose pre-image, which is analyzed in real time and then used to set the optimum kVp and mAs for the exam. Information on breast density also can be measured and used in the AEC algorithm.

Output Devices

Film images are read as hard copy, viewed on a light box. Initially, digital mammography images have been viewed the same way. The digital image is captured and stored on a computer, then printed on a high-definition laser printer to be viewed as film.

In the near future, soft-copy review via workstation displays will offer the full benefits of digital imaging. Digital transmission, easy retrieval and display of previous images, and on-line image processing (window-level control and feature magnification) are all advantages of digital imaging. With images of 30 million pixels or more, digital mammography places great constraints on the monitor performance. However, resolution, speed, and light-output characteristics of display monitors are approaching the requirements needed for digital mammograms.

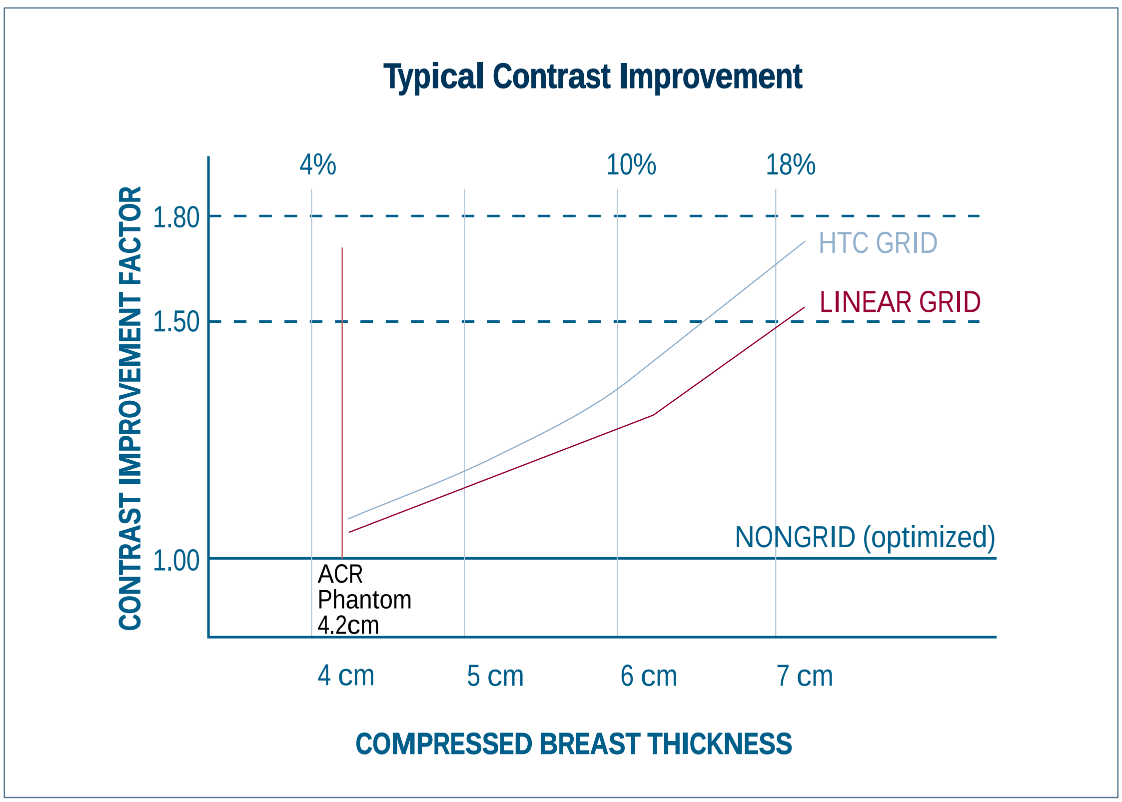

Figure 14. Contrast improvement for linear and high transmission cellular grids.

Future Applications

Digital mammography offers some important future applications that are not practical or possible with screen-film. The clinical utility of these applications is under active investigation.

Stereotactic Breast Biopsy

A full-field digital mammography detector can also be used for stereotactic breast biopsy imaging. This enables a multi-purpose system ideal for both screening and follow-up imaging tasks.

Stereo Mammography

Acquisition of two images, separated by a few degrees, allows the display of mammograms in a stereo fashion to facilitate the visualization of three-dimensional characteristics. Computer display of stereo images requires either synchronized glasses to display frame updates or polarizing glasses. Investigations of this technique[10] show promise in improving perception in mammography.

Tomosynthesis and Three-Dimensional Imaging

These applications utilize acquisition of multiple images from many angles, with the data processed to generate tomographic slices. These images will exhibit reduced confusion from overlapping structures. Preliminary studies[11] show that these acquisitions can be performed with total doses similar to screening mammography. The resulting images offer reduced overlap and better diagnostic information not available with planar imaging. The high quantum efficiency of selenium direct detectors will be important for these applications in reducing the total dose and maximizing image quality for the multiple images required. Large field-of-view detectors are also important to ensure the acquisition of full-field images that do not truncate at the field-of-view edge and create artifacts.

The requirements for a successful tomosynthesis mammography detector are:

- High DQE, for improved image quality in the reduced-dose subimages
- High DQE at high x-ray energies, allowing use of higher kV and reduced breast compression
- Large field of view to avoid cutoff at breast edges and potential reconstruction artifacts

Dual-Energy Imaging

In dual-energy imaging, two images are made of the breast at different x-ray energies. Because of the differing x-ray attenuation characteristics of glandular tissue, adipose tissue, and microcalcifications, processing of the dual images can enhance the visibility of certain structures. The clinical utility of this has been suggested,[12] but further research is needed. Dual-energy imaging can also be used to quantify fibroglandular breast tissue density.

Computer-Aided Detection

The use of computer programs to perform detection is an area of active research.[13, 14] The digital image is examined by software, and suspicious areas are highlighted for further scrutiny by the radiologist. The challenge for these systems is finding the proper balance between sensitivity and specificity. Increasing sensitivity can result in too many false positives marked on an image, and the radiologist may find this more of an annoyance than a help. Conversely, if not enough true positives are marked, the system will be offering little help to the radiologist. The low cancer rate per image in a screening environment makes this task challenging. However, these types of programs eventually will become routine, reinforced by the convenience of performing CAD procedures on digitally acquired images.

Digital mammography offers many potential advantages over film for CAD systems. Digitizing of the film images is time consuming and yields a less-than-optimal digital image. True digitally acquired images are expected to improve the accuracy of CAD systems, because these images offer an increased dynamic range and access to multiple views and previous studies.

Conclusions

Amorphous selenium detectors represent a new standard for mammography systems. Amorphous selenium is the ideal material for digital mammography systems due to its high intrinsic resolution and direct-conversion abilities. Selenium detectors outperform both film and indirect digital detectors in terms of resolution, signal-to-noise ratio, and dose efficiency. In addition to the communication and storage efficiencies of a digital environment, these direct conversion mammography systems offer the potential for greater image quality without increasing radiation dose.

Glossary

a-Se	Amorphous selenium. The most common photoconductor material used in direct detectors.
CsI(Tl)	Cesium-iodide with thallium doping. A common scintillator used in indirect detectors.
Direct Conversion	A method of detecting X-rays utilizing a material that directly absorbs an X-ray and generates an electrical signal.
DQE	Detective Quantum Efficiency. Measure of the square of the output signal-noise ratio to the input signal/noise ratio, as a function of spatial frequency.
FFDM	Full-Field Digital Mammography. Digital systems offering field of view large enough to image the entire breast.
Gd_2O_2S	Gadolinium oxysulfide. A common phosphor used in intensifying screens and some indirect detectors.
HTC Grid	High Transmission Cellular grid. Anti-scatter grid design, using a cellular grid construction with air as the interspace material.
Indirect Conversion	A method of detecting X-rays requiring a two step process using a scintillator to generate light photons upon absorption of X-rays, and a light-sensing element to convert the light photons into an electrical signal.
MQSA	Mammography Quality Standards Act. Defines minimum quality standards for mammography equipment, facilities, and operators.
MTF	Modulation Transfer Function. Measure of the system contrast response, as a function of spatial frequency.

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