Fundamentals of Digital Mammography:

Physics, Technology and Practical Considerations

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Introduction

Screen-film image receptors have been the standard detector used in conventional mammography. New developments in detector technology and computers are altering the landscape of mammography imaging. Full Field Digital Mammography (FFDM) offers the promise of revolutionizing the practice of mammography through its superior dose and contrast performance. Advanced applications made possible through digital imaging, such as automated computer-aided diagnosis, dual-energy and 3D tomosynthesis are expected to further improve diagnostic sensitivity and specificity.

This primer describes the technical basis for current and future advances in mammographic detector technology. These include all of the following:
- Lower dose
- Improved image quality
- Computer-aided diagnosis
- Softcopy review and digital archiving
- Tomosynthesis and other three-dimensional visualization techniques
- Reduction in breast compression pressure

New flat-panel X-ray detectors offer extremely high quantum efficiency and high resolution. These will translate into lower dose and improved image quality mammograms. Digital detectors for mammography can be categorized as indirect or direct conversion detectors. This primer describes these technologies and how direct conversion detectors, in particular selenium-based detectors, provide the best performance for full field digital mammography systems.

Indirect conversion digital detectors utilize a method of imaging X-rays, similar to screen-film, wherein a scintillator absorbs the X-rays and generates a light scintillation. The light scintillation is then detected by an array of photodetectors. However, indirect conversion systems suffer from resolution degradation caused by light spread in the scintillator, and from poor quantum efficiency caused by the use of thin scintillators.

Direct conversion 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 the most advanced direct conversion technology for digital mammography. Selenium is an ideal material for a mammography detector, because it has high X-ray absorption efficiency (approaching 100 percent at mammographic X-ray energies), extremely high intrinsic resolution, low noise and a well-established manufacturing process.

The pixel size requirement for FFDM is also addressed in this article. Light blurring in indirect conversion systems interferes with the effective design and performance of systems with very fine resolution. Direct conversion systems, in contradistinction, can have their resolution characteristics optimized independently from their quantum detection efficiency and can thus be designed with an optimum pixel size.

The field of view for FFDM systems is very important. In order to be able 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 commonly used in screen film imaging, 24 x 30 cm. Detectors that are the size of the smaller cassette, 18 x 24 cm, will require imaging and tiling of many images to cover the breast field of view for larger breasts. For the U.S. demographics, up to 30 percent of all women require the larger field of view.

Detector performance is commonly quantified by two metrics: Modulation Transfer Function (MTF) and Detective Quantum Efficiency (DQE). MTF is a measure of resolution and DQE is a measure of dose efficiency. The MTF for screen-film is superior to indirect conversion detectors, while the MTF for direct conversion detectors are superior to both. The DQE of indirect conversion detectors is superior to screen-film, however the DQE of selenium direct conversion detectors is seen to be superior again to both screen-film and indirect conversion detectors. With superior DQE at all spatial frequencies, selenium detectors offer the potential for both improved image quality and lower patient dose.

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

Systems with high quantum efficiency, especially at increased X-ray energies, offer the possibility of decreased breast compression. As is commonly known, painful breast compression is an important factor in undercompliance of women to present for routine breast screening.

Future applications such as stereo mammography, breast tomosynthesis, contrast enhanced digital mammography and other 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 selenium detectors offer. The digital detectors will also be able to be used for full-field and high resolution stereotactic breast biopsies and diagnostic imaging tasks. It is essential, however, that the full-field digital systems perform well for its primary task, i.e. breast screening.

In conclusion, digital detectors offer many advantages compared to screen-film imaging for mammography. It is hoped that the potentials for improved image quality, lower dose and advanced imaging applications will result in improved diagnostic accuracy.

Screen-Film Mammography

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

An important parameter to understand is the thickness of the intensifying screen. Thicker screens capture more X-rays and are therefore more dose efficient and higher speed. However, thicker screens also create more light scatter and blurring of 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.
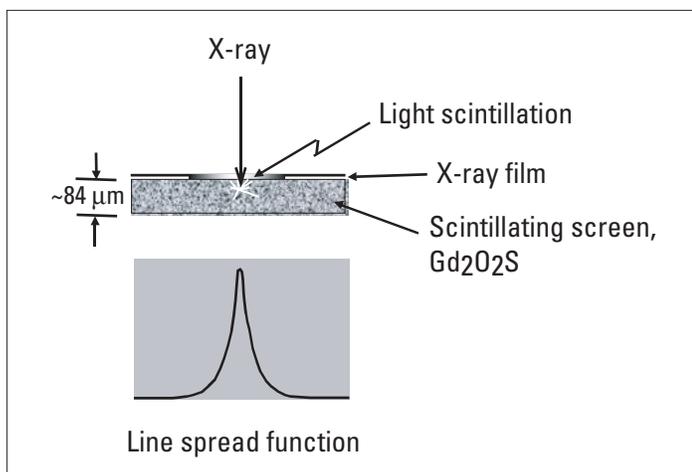
Figure 2 shows the performance trade off of sensitivity versus resolution inherent in the design of screen-film systems. As the screen is made thicker, the cloud of light on the film will increase in size, on average. This reduces the resolution of the system, however the system's sensitivity increases because the thicker phosphor increases the probability that the incoming X-ray will be absorbed. Thus screen-film systems have a performance trade off between speed and resolution.

Because X-rays are absorbed with a decaying exponential spatial distribution, in a screen-film mammography system the film is placed at the entrance surface of the scintillating screen.

While a screen-film system offers several advantages, there are significant disadvantages and limitations of this system. Film does not have a linear sensitivity to photonflux, and there is a narrow range over which it can detect small differences in contrast. In particular, tissue areas of high and low density are often suboptimally imaged. Frequently, the entire image is poorly exposed because of film's stringent requirements for proper exposure, resulting in repeated imaging.

Another major problem is film granularity, affecting detective quantum efficiency at high optical densities and visibility of microcalcifications.

Film also requires processing time and storage space. Chemical processing of the film 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 detectors. Because images are captured as a digital signal, 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
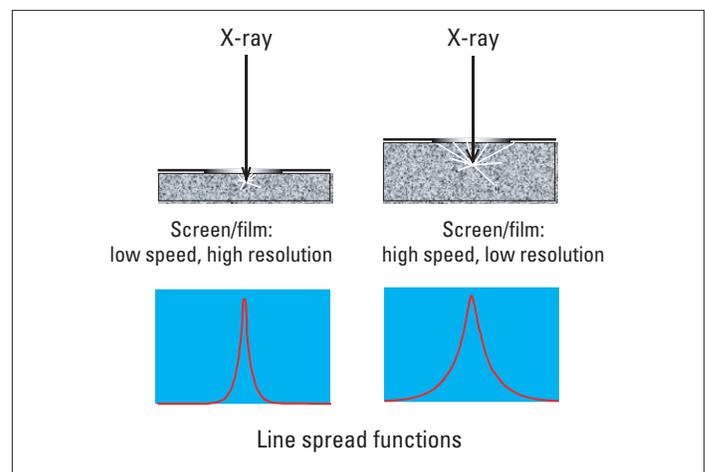

Figure 1.
Screen-film systems use a scintillating screen to absorb the X-ray and generate light photons, which are captured on the film.

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

adjustment to optimize contrast for every imaging task. Soft-copy reading, computer-aided diagnosis and three-dimensional imaging offer additional and potentially 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 conversion and direct conversion.

Indirect-Conversion Digital Detectors

The earliest digital mammography systems available in the U.S. used indirect conversion detectors. Such detectors use a two-step process for X-ray detection, as seen in Figure 3. The first step requires a scintillator layer 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 converts light photons to electronic signals that are captured using thin-film transistors. Some systems, like Charge-Coupled Devices (CCDs), use alternative light collection and readout methods. In both systems, the light sensitive imaging function of film has been replaced by digital light imaging. It is in this sense that these can be seen as an evolution of screen film imaging.

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

The placement of the scintillator is more problematic in indirect conversion digital detectors than with screen-film systems. As with film screens, 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 transpare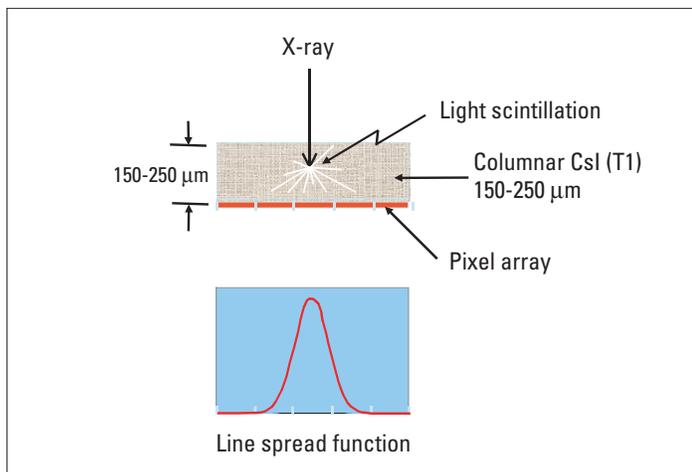nt to X-rays and the array must be placed on the exit surface of the scintillator. This causes degradation in spatial resolution compared to screen-film.

Typical thicknesses of CsI(Tl) used in mammography detectors range from 150 to 250 microns, and these indirect conversion digital detectors exhibit light spreading similar to screen-film systems. Examples of indirect conversion detectors are the Fischer Imaging CsI/CCD-based detector and the GE CsI/TFT detector.

Direct-Conversion Digital Detectors

Direct-conversion digital detectors represent a technological advance, eliminating problems associated with light scatter inherent in indirect conversion systems. In these systems, illustrated in Figure 5, a photoconductor absorbs the X-rays and directly generates the signal (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 the direction of the electric field lines, they move without lateral charge spreading. This results in an exceptionally narrow point spread response, of about 1 micron.

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

In direct-conversion 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. This is shown in Figure 6. In practice, the photoconductor is made sufficiently thick in order to stop
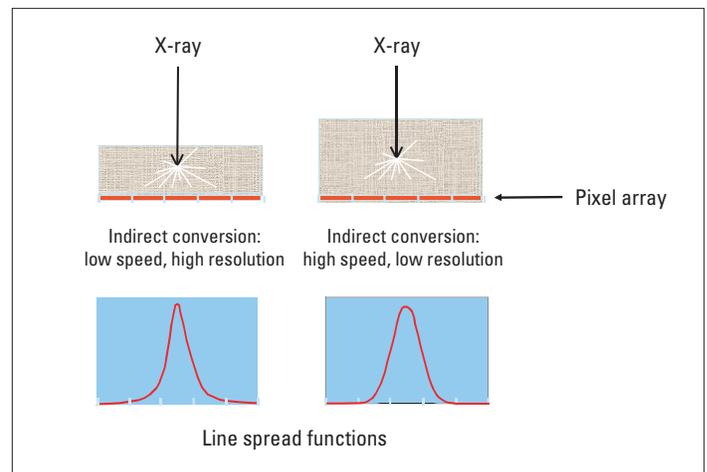

Figure 3.
Indirect conversion detectors utilize a scintillating layer to absorb the X-ray and generate light photons, which are detected by a photodiode array.

Figure 4.
Indirect conversion detectors generate poorer resolution images as the scintillator thickness is increased.
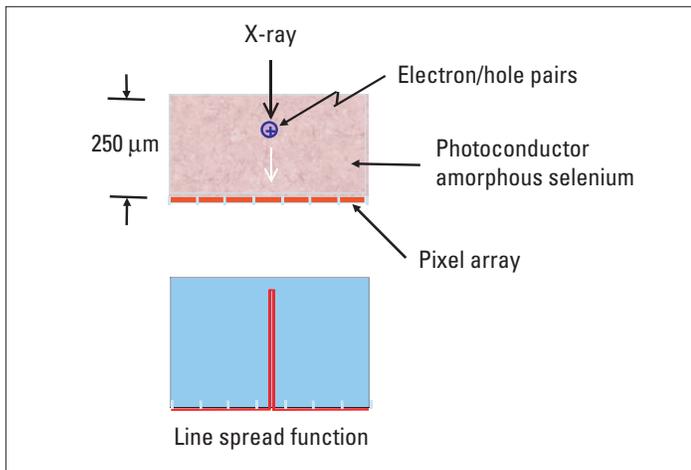
Figure 5.
Direct-conversion detectors use a photoconductor to absorb the X-ray and directly generate the signal.
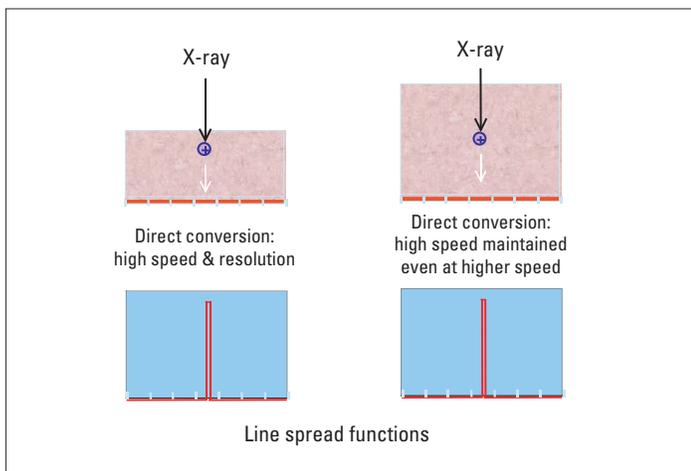

Figure 6.
There is no tradeoff between spatial resolution and sensitivity for direct-conversion systems.
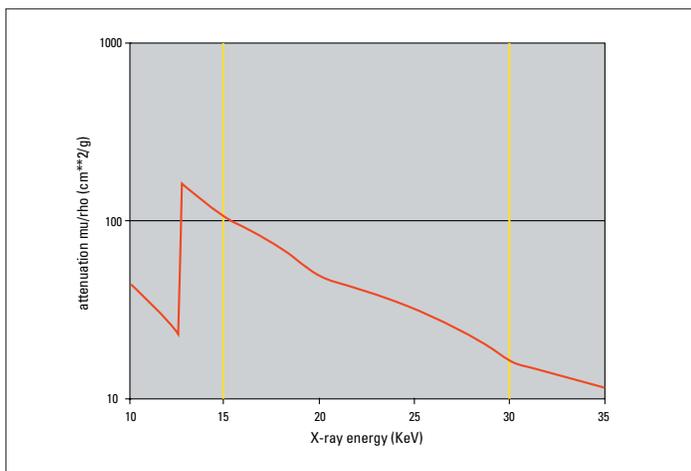

Figure 7.
K-edge for Selenium is just below the diagnostic range for mammography, shown in the area between the two yellow lines.

the majority of the incident X-rays, and this can be done without adversely affecting the spatial resolution, an important consideration in mammography with its dual needs of high resolution and low radiation exposure.

Using amorphous selenium as the photoconductor, a thickness of 250 microns is adequate to stop more than 95 percent of the X-rays in the mammographic energy range. This is seen in Figure 7. Standard screens for use in film mammography only have about 50 to 70 percent quantum efficiency, and the scintillator CsI(Tl) used in indirect-conversion digital detectors exhibits about 50 to 80 percent quantum efficiency. Systems using amorphous-selenium can achieve almost complete quantum efficiency.

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.

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

Semiconductor arrays for direct conversion detectors are much easier to fabricate than arrays for indirect conversion detectors because selenium-based detectors do not require a photodiode structure on top of the TFT. As a result, these detectors can utilize the same manufacturing processes as large area TFT-LCD arrays, commonly used in computer displays and do not require construction of dedicated manufacturing facilities.

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 very close together on the semiconductor surface. These systems use fiber optics to capture images from light emitted from scintillators or intensifying screens, but suffer from light loss and added complexity due to the fiber optics. The only commercially available CCD-based system uses a slot scanning technology, whereby a small detector is scanned across the breast to build up the image over several seconds. This is in distinction to the other FFDM systems that acquire the image in a single snapshot.
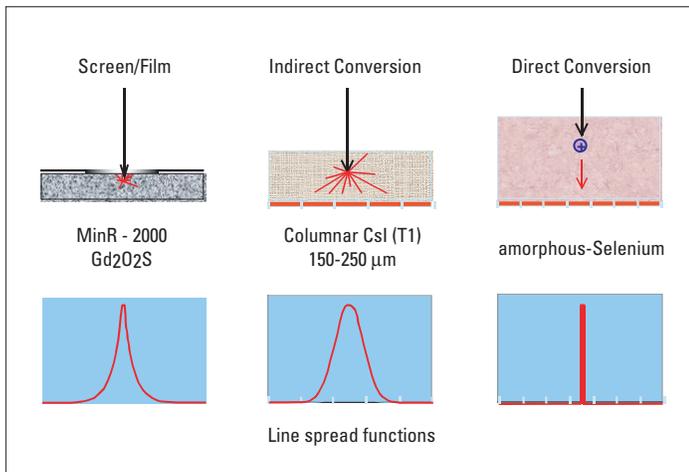
Figure 8.
Line spread functions for screen-film and digital detectors. Screen-film and indirect-conversion detectors have broad presampled line spread functions; direct-conversion detectors have narrower presampled line spread functions.

Pixel Size Requirements

Mammographic imaging requires the detection and classification of extremely small objects. In particular, microcalcifications can be as small as 100 to 200 microns. Any useful FFDM system must be able to image these smallest microcalcifications of interest.

Digital detectors are comprised 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. Figure 8 shows the intrinsic resolution for direct and indirect detectors.

In indirect-conversion digital detectors, as the pixel size is decreased, a limit is reached beyond which further reductions in pixel size does not improve resolution. This resolution limit is a function of the light scattering in the scintillator. This limit is about 100 microns in practical indirect conversion scintillators, which is the upper limit for application to digital mammography. CCD based indirect conversion detectors typically have much smaller pixels, however the light spread from the scintillator limits the resolution so the result is an image with more pixels but not superior resolution.

In a direct-conversion digital detector, on the other hand, spatial resolution is only limited by the size of the pixel. The size of the pixel in these detectors can be made arbitrarily small (within limits) to make the resolution performance extend to very high spatial frequencies. The ultimate limit to very small pixels is the reduced X-ray flux impinging upon the detector.

The pixel size for FFDM systems range from about 50 to 100 microns. As explained earlier, the indirect-conversion detectors all have resolution characteristics that are scintillator limited, so the raw pixel size does not accurately reflect the actual resolution characteristics of the image. The pixel size for the Hologic's selenium detector is 70 microns, and because of the design of this detector, this represents its true resolution characteristic.

Pixel size design represents a compromise between spatial resolution and system cost. As the pixel size of any digital imaging system is made smaller, the amount of data contained in the image rapidly increases, which increases system costs in terms of data storage, network bandwidth and display capabilities. Fabrication of large-area detectors with very small pixels is an expensive, low-yield process, and integration of such detectors with analog and digital readout electronics and system interconnects pushes the state of the art in microfabrication on several fronts.

Field of View Requirements

MQSA requirements for mammography facilities require the use of 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. On the other hand, the use of a large cassette on smaller women is wasteful of film and processing materials.

Digital mammography presents additional field-of-view issues. At the present time, flat-panel X-ray detectors are considerably more expensive than screen-film cassettes, precluding the use of more than one receptor size on a given machine. Irrespective of cost, digital detectors are heavy and fragile and do not lend themselves to rapid changeover as do film cassettes. Therefore, digital mammography equipment manufacturers must decide on one optimum detector size, one that is large enough to accommodate most women without requiring double exposures.

A digital detector of size 18 x 24 cm (the smaller film cassette size) is inadequate to image approximately 20 to 30 percent of U.S. women. Because of regional differences, some areas of the country might have a larger percentage of women that cannot be screened using the smaller film cassette.

If the breast is too large to image on the detector in one exposure, multiple exposures that 'tile' the breast are required. This solution has disadvantages such as requiring multiple additional compressions, additional setup and imaging times, breast regions that suffer from repeated radiation exposure, and the difficulty of radiologist's review of multiple images.

Use of a large detector for digital mammography is an important design issue. Issues using a large digital detector to image a small breast does not present the same problems as film. The large image can be cropped to reduce data transfer, display and storage space requirements, and there are no additional material costs. The one technical challenge in the use of a large detector for small breasts is in positioning, however the use of smaller compression paddles in conjunction with the large detector solves this problem.

The larger detector can be used to advantage when imaging smaller breasts. The imaging can be performed in a magnification mode, creating the effect of smaller pixels and reducing scattered radiation at the same time. If a system is designed so that breasts

that are adequately served by the smaller film cassette size of 18 x 24 cm are imaged in magnification mode to project onto a detector of size 24 x 30 cm, an image of effective pixel size of 56 microns will be achieved for systems with inherent 70 mm pixel sizes. Such a system could offer 56 micron imaging in a screening environment for approximately 80 percent of U.S. women. This is another possibility offered by use of a large digital detector.

Characterizing Detector Performance

There are parameters that characterize the overall image quality obtained on mammography and other radiography systems. The most important are:
- Spatial resolution
- Contrast resolution
- Signal-to-noise ratio
- Dose efficiency

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 9 shows the need for improved dynamic range in breast imaging. Digital detectors offer greatly improved performance. It has been shown that for an ideal detector with no inherent detector noise, 3100 gray levels are discernable in a typical mammographic image. Thus a system that provides at least 12 bits of dynamic range will not degrade the underlying information. Digital mammograms will
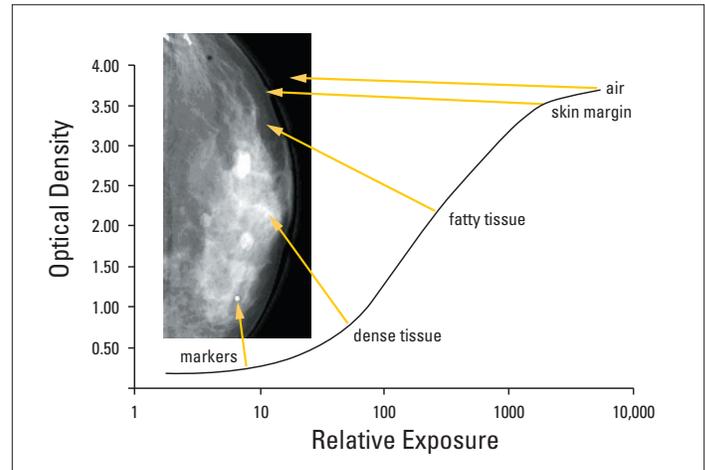

Figure 9.
Characteristic curve of mammographic film illustrating how the display contrast (slope of the curve) is suboptimal in lucent and dense regions of the breast[1].

exhibit uniform quality over a range of exposure conditions and breast sizes.

X-ray Quantum Detection Efficiency

X-ray quantum detection efficiency is a measure of the percentage of X-rays that hit the detector that are absorbed. Systems with higher quantum efficiency can produce higher quality images, at lower dose. As mentioned earlier, selenium systems stop more than 95 percent of the X-rays in the mammographic energy range. Standard screens for use in film mammography only have about 50 to 70 percent quantum efficiency, and CsI(Tl) used in indirect conversion detectors exhibit about 50 to 80 percent quantum efficiency.

Modulation Transfer Function (MTF)

Modulation Transfer Function (MTF) and Detective Quantum Efficiency (DQE) provide quantitative measurement of imaging performance. MTF measures spatial resolution, while DQE is a measure of signal-to-noise ratio, contrast resolution and dose efficiency. An imaging system is best characterized by examining corresponding MTF and DQE curves; however, it 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) is a measure of signal transfer over a range of spatial frequencies and quantifies spatial resolution. Spatial resolution is often measured using a line pair phantom, illustrated in Figure 10. The ultimate resolution limit of any system is determined by its pixel size. For example, a system with a 100-micron pixel cannot adequately resolve spatial frequencies above 5 line pairs/mm (lp/mm). Indirect-conversion methods can scatter light over several pixels, 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.

Commercially Available Digital Mammography Detectors

The following is a list of full-field-of-view digital mammography detectors commercially available in the U.S.

Indirect–conversion detectors
- GE
Scintillator	CsI(Tl)
Pixel size	TFT 100 microns
Field of view	18 x 23 cm

- Fischer Imaging
Scintillator	CsI(Tl)
Pixel size	CCD 24/48 microns
Field of view	22 x 30 cm (scanning)

Direct–conversion detectors
- Hologic/Lorad
Photoconductor	amorphous selenium
Pixel size	TFT 70 microns
Field of view	24 x 29 cm

As shown in Figure 11, the MTF of a selenium direct conversion detector[2] is superior to those of screen-film[3] and indirect-conversion detectors[4].

The intrinsic spatial resolution of selenium direct-conversion mammography detectors is superior to those using indirect-conversion scintillators. While MTFs for indirect conversion detectors fall dramatically at higher spatial frequencies, the MTF of direct-conversion selenium detectors remain 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. Detective Quantum Efficiency (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.

Figure 12 shows the DQE for film and digital detectors. While film has a high MTF at high spatial frequencies compared to the indirect conversion detectors, the same is not true for DQE. 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.

Indirect conversion systems have a superior DQE to screen-film, especially at the lower spatial frequencies. However, the DQE drops at high spatial frequencies—a consequence of the scintillator-induced light blurring.

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

Direct-conversion systems offer the possibility of similar image quality at lower doses (by as little as 1/2 to 1/3 of the screen-film dose) 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 conversion FFDM systems may provide superior diagnostic imaging in terms of sensitivity and specificity, resulting in better cancer detection and lower false positive rates.
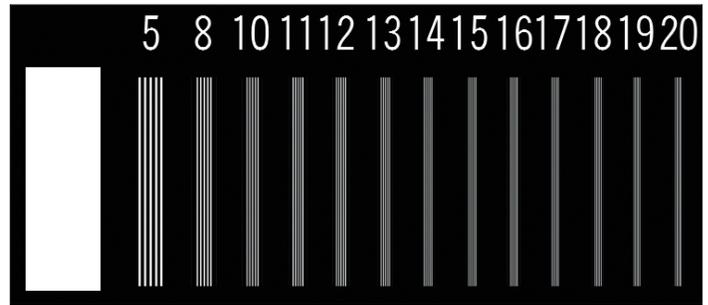

Figure 10.
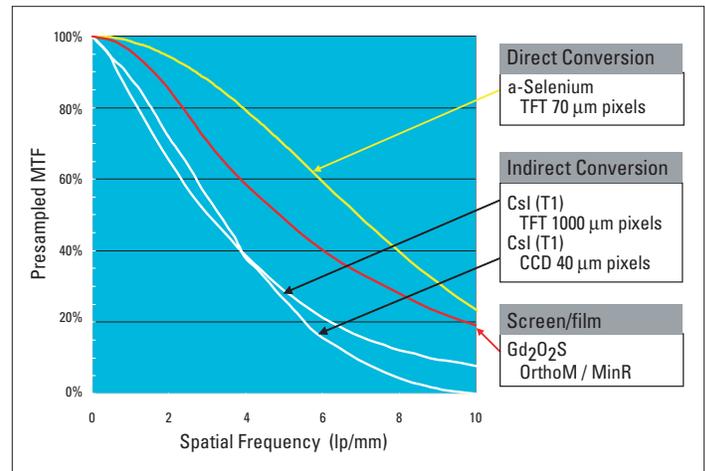X-ray resolution test pattern to determine imaging resolution performance, in units of line pairs/mm.

Figure 11.
Modulation Transfer Function for screen-film (Kodak: OrthoM/MinR), indirect conversion and direct conversion digital mammography detectors.
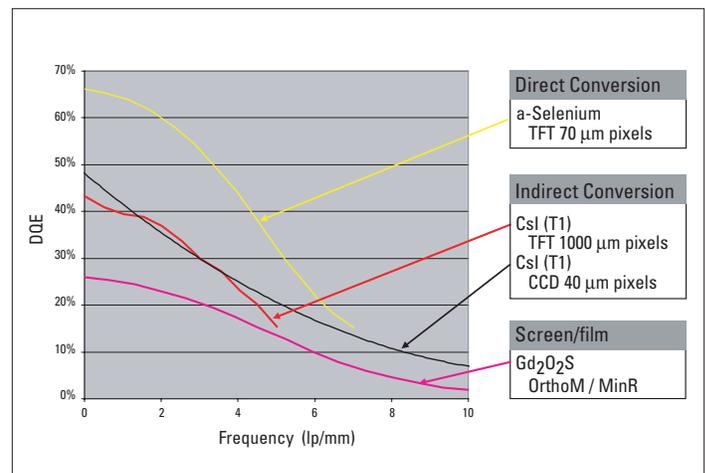

Figure 12.
Detective Quantum Efficiency for screen-film (Kodak: OrthoM/MinR), indirect-conversion and direct-conversion digital mammography detectors.
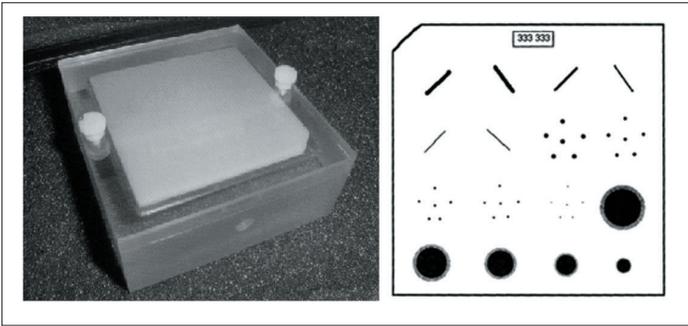
Figure 13.
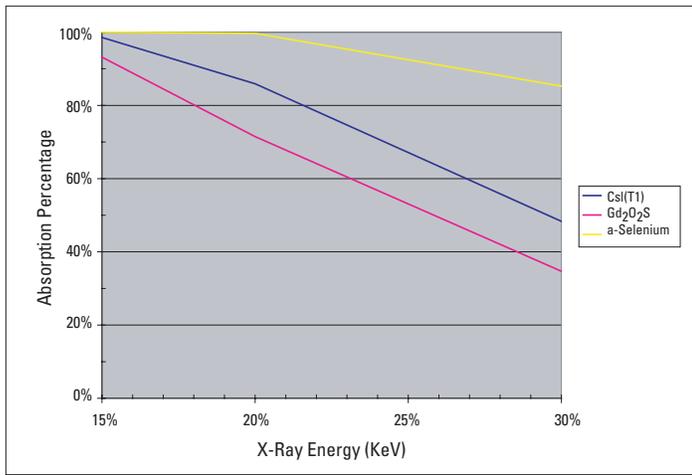ACR Mammography Accreditation Phantom, and imaging details.

Figure 14.
Percent absorption of incident X-rays for materials used in screen-film: Gd_2O_2S 34 mg/cm², indirect conversion CsI(Tl) : 73 mg/cm² and second generation cselenium: 250 microns thick.

ACR Phantom Imaging Performance

Figure 13 shows the acrylic phantom used for ACR accreditation and MQSA inspections. The phantom is approximately equivalent in X-ray absorption to a 4.2-cm thick compressed breast consisting of 50 percent glandular and 50 percent adipose tissue. The phantom includes appropriate details that range from visible to invisible on a standard mammographic film image. The phantom has fibers with diameters of 1.56, 1.12, 0.89, 0.75, 0.54 and 0.40 mm; specks with diameters of 0.54, 0.40, 0.32, 0.24 and 0.16 mm; and masses with decreasing diameters and thicknesses of 2.00, 1.00, 0.75, 0.50 and 0.25 mm.

The visibility of phantom details has been evaluated for screen-film and for first and second-generation detectors.[5] The phantom was imaged at 28 kVp at a dose of ~3 mGy. (A high dose was used to emphasize the performance differences.) Results are summarized in Table 1. These results are consistent with predicted performance based on the DQE plots: second-generation systems outperform both screen-film and indirect conversion.

System Design Considerations

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

X-ray Source

The spectrum of X-ray energies used in screen-film mammography has been highly optimized. Digital mammographic detectors offer improved performance characteristics, particularly dynamic range, however the issue of the optimum energy is still in the investigational phase.[6] It is possible that higher X-ray energies may permit lower dose or higher image quality with digital mammography, particularly for patients with dense breasts. Detectors that have high intrinsic quantum efficiency allow the use of higher energies.

Selenium has superior stopping power compared to film screens and CsI(Tl) material, as seen in Figure 14. 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. Systems that reduce breast compression will utilize higher X-ray energies than currently used, and the pronounced high-energy performance of selenium will become even more important.

Scatter Rejection Methods

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

Anti-scatter Grids

There are two common methods of grid design, shown in Figure 15. Standard linear grids are constructed of long thin slats of radio-opaque materials (called septa or laminae) separated by

Detector	Technology	Fibers visible	Specks visible	Masses visible
Kodak MinR 2000	Screen film	4	4	3
CCD	Indirect conversion	5.5	4	5
a Selenium	Direct conversion	6	5	5

Table 1.
Performance of systems in imaging the ACR Mammography Accreditation Phantom

Figure 15.
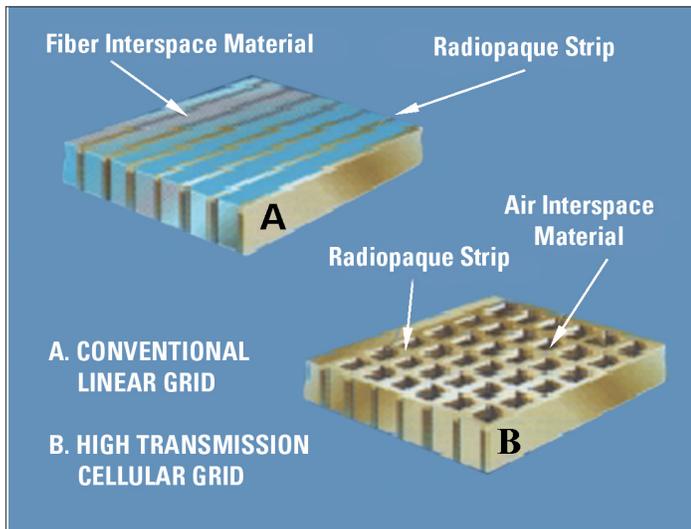Construction of linear and cellular grids.

Figure 16.
Contrast improvement for linear and high transmission cellular grids.

relatively radiolucent spacer materials. These grids effectively reduce scattered photons, but also block some of the primary beam, mainly from absorption in the spacers. An alternative method of grid construction utilizes a crossed air grid, known as a High-Transmission Cellular (HTC) grid. This grid absorbs scattered radiation in two directions, as opposed to the linear grid, which removes scattered radiation only in one direction. This grid absorbs less of the primary beam, because the intersepta material is air.

The HTC grid has been shown to have superior contrast improvement (scatter rejection) and Bucky factors (primary beam reduction) to conventional linear grids.[7] For a 7-cm compressed breast, the HTC grid offers a contrast improvement of about 1.7, compared to a linear grid having a contrast improvement of about 1.5.

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 much wider dynamic range and consequently are more tolerant of exposure variations. Therefore, digital mammography should have a reduced number of retakes compared to film-screen. 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 can also be measured and used in the AEC algorithm.

Breast Compression

The need for compressing the breast during mammography imaging will not be eliminated with the use of digital detectors.
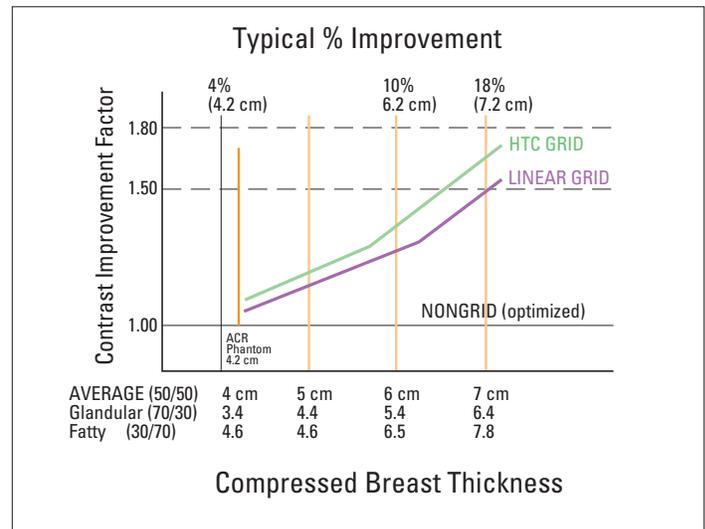
The relatively large compression pressure used in screen-film mammography is in part due to the limited dynamic range of film, and the use of low energy X-rays. FFDM systems have dynamic range superior to film. Selenium detectors have high quantum efficiency even at higher X-ray energies than used today, with the potential of obtaining high quality images with less compression. Large breast compression is used for other reasons, among them reducing overlap of breast structures. This is an area that requires additional study.

Output Devices

Film images are read as hard copy, with the images 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 analogously to film.

Soft copy review via workstation displays offers the true benefits of digital imaging. Digital transmission, easy retrieval and display of previous images and on-line image processing such as windowing, leveling and feature magnification are all advantages of a digital image. Digital mammography places great constraints on the monitor performance, however. The images are large (~10,000,000 pixels, or more) but resolution, speed and light-output characteristics of display monitors are approaching the requirements needed for digital mammograms.

Future Applications

Digital mammography offers some exciting 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 diagnostic and stereotactic-breast-biopsy imaging. This enables

a multi-purpose system ideal for both screening and follow-up imaging tasks.

Stereo Mammography

Acquisition of binary images, separated by a few degrees, allows the display of mammograms in a stereo fashion, to facilitate the visualization of the three-dimensional characteristics of mammographic features. Computer display of stereo images requires either glasses synchronized to display frame updates, or polarizing glasses. Investigations of this technique[8] show promise in improving three-dimensional depth discrimination in mammograms.

Tomosynthesis and Three-Dimensional Imaging

These applications utilize acquisition of multiple images from many angles, typically 9-15 about a +- 15 degree range, with the data processed to generate tomographic slices. These images will exhibit reduced confusion from overlapping structures. Preliminary studies[9] show that these acquisitions can be performed with doses similar to screening mammography, and that the images offer diagnostic information not available with 2D imaging. High quantum efficiency will be important for these applications, to reduce the total dose for the multiple images required.

Dual-Energy Imaging

In dual-energy imaging, two images are made of the breast at differing 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[10], but further research is needed. Dual energy imaging can also be used to quantify fibroglandular breast tissue density.

Computer-Aided Diagnosis

The use of computer programs to perform preliminary cancer diagnoses is an area of active research.[11] The digital image is examined by software, and suspicious areas are highlighted for further scrutiny by the radiologist. The challenge for these systems is to find 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 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 difficult. However, it is expected that eventually these types of systems will become routine, since it is very easy to perform the CAD procedure on the digitally acquired image.

Digital mammography CAD systems offer significant advantages compared to film-based CAD systems. Digitizing of the film images is time consuming and yields a less than optimal digital image. With a true digital image, the increased dynamic range and access to multiple views and previous studies is expected to improve the accuracy of CAD systems.

Conclusions

Full field digital mammography offers the potential for significant advances in breast cancer diagnosis, including lower radiation dose, reduced breast compression pressure and improved cancer detection rates. FFDM has advanced rapidly in the past few years, and there are now several systems commercially available for sale in the U.S. These systems offer improvements over screen-film imaging, especially in regards to imaging dynamic range and digital storage and display. The different manufacturer's products can be differentiated by the image receptor's detection technology, its resolution dose and scatter rejection performance, and in the detector's field of view.

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Glossary

a-Se . . . . Amorphous selenium. The most common photoconductor material used in direct conversion detectors.

a-Si . . . . Amorphous silicon. The material used to manufacture TFT arrays used in both indirect and direct FFDM image receptors.

CsI(Tl) . . . . Cesium-iodide with thallium doping. A common scintillator used in indirect conversion 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 film screens and some indirect conversion detectors.

HTC Grid . . . . High Transmission Cellular grid. Anti-scatter grid design, using a cellular grid construction with air as the inter-septa material.

Indirect Conversion . . . . A two step method of detecting X-rays. First, a scintillator is used to generate light photons upon absorption of X-rays, then a light sensitive element is used to convert the light photons into an electrical signal.

MQSA . . . . Mammography Quality Standards Act, that defines minimum quality standards for mammography equipment, facilities and operators.

MTF . . . . Modulation Transfer Function. Measure of the system response, as a function of spatial frequency.

Scintillator . . . . A material that gives off light photons upon absorption of an X-ray. Used in screen-film and indirect conversion digital imaging.

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