Digital Mammography1

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2 authors:

Etta Pisano
Medical University of South Carolina

Martin J Yaffe
University of Toronto

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Stephen A. Feig, MD • Martin J. Yaffe, PhD

■ INTRODUCTION

The successful use of screen-film mammography in breast cancer screening is one of the major achievements of medical imaging in this century. However, screen-film mammography also has a number of limitations, a fact that has encouraged investigation of whole-breast digital mammography as a possible alternative. In this article, the achievements and limitations of conventional mammography as well as the potential advantages and applications of digital mammography are discussed. The article also describes digital mammography systems currently being developed, including potential limitations and design and cost considerations. In addition, possible future developments are discussed.

■ ACHIEVEMENTS OF SCREEN-FILM MAMMOGRAPHY

Statistically significant reductions in breast cancer mortality have been realized at five randomized controlled trials of screening mammography in Sweden, including a reduction of 34% among women aged 50–74 years at the time of entering the Swedish Two-County Trial (screening offered every 33 months) (1), 36% among women aged 45–49 years at the time of entering the Malmö trial (screening offered every 2 years) (2), and 45% among women aged 39–49 years at the time of entering the Gothenburg trial (screening offered every 18 months) (3). A metaanalysis of the combined results for women aged 39–49 at the time of entering the five Swedish trials showed a 29% reduction in breast cancer mortality with screening mammography offered at intervals varying from 18 to 28 months (4). Moreover, it has been suggested that substantially greater reductions in mortality would have resulted if screening mammography had been offered annually or if all women had accepted the offer of screening (5–7).

Although no randomized controlled trials with modern mammography have been conducted in the United States, two major North American service screening programs that use modern mammography have generally enabled earlier detection of breast cancers than the randomized controlled trials (8). These results have been achieved while maintaining acceptably low rates of recall for additional imaging after the basic screening examination (10% for the initial prevalence screening and 5% for subsequent incidence screenings) as well as maintaining false-positive biopsy rates for nonpalpable lesions that are lower than those for lesions found at clinical examination (9). Positive predictive values for lesions for which biopsy is recommended on the basis of mammographic findings range from 25% for women in the 5th decade of life to 50% for women in the 8th decade of life and are all within an acceptable range. Rates for interval cancers that surface between annual screenings are low, ranging from 10% to 20% (10).

These remarkable achievements have been made possible by the technical capabilities of screen-film mammography. These capabilities include high resolution, which demonstrates fine spiculations and microcalcifications; high contrast, which allows visualization of subtle differences among soft-tissue densities; use of high-luminance view boxes, which improve visualization of dense tissue; ease of simultaneously displaying, rearranging, and masking the basic four-image screening examination along with supplementary views and previous images on a multiple panel illuminator; and the availability of both 18 × 24-cm and 24 × 30-cm film, enabling the imaging of breasts of different sizes.

Abbreviations: CAD = computer-aided diagnosis, H&D = Hurter and Driffield

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[1]From the Breast Imaging Center, Department of Radiology, Thomas Jefferson University Hospital, 1100 Walnut St, Philadelphia, PA 19107-5563 (S.A.F.); and the Department of Medical Biophysics, Sunnybrook Health Science Centre, University of Toronto, Toronto, Ontario, Canada. From the Plenary Session, Friday Imaging Symposium: Advances in Women's Imaging, at the 1997 RSNA scientific assembly. Received March 13, 1998; revision requested April 7 and received April 14; accepted April 14. **Address reprint requests to** S.A.F.

■ LIMITATIONS OF SCREEN-FILM MAMMOGRAPHY

Despite its numerous advantages, screen-film mammography also has limitations, one of which results from the trade-off between dynamic range (latitude) and contrast resolution (gradient). The relationship between x-ray exposure, image density, and contrast is illustrated by the sigmoid Hurter and Driffield (H&D) curve that is characteristic for a given type of screen-film system under specific conditions. Because of the limited range of soft-tissue densities in the breast, mammography requires high contrast. The fixed characteristics of an H&D curve mean that if high contrast is to be obtained in intermediate-density tissue, there must be lower contrast within the thicker, denser fibroglandular tissues represented at the toe of the curve and the fatty tissue represented at the shoulder of the curve (Fig 1). Screen-film mammography will typically have a much lower dynamic range than chest, bone, and gastrointestinal radiography.

Another limitation of screen-film mammography is the necessary trade-off between the x-ray utilization efficiency of the screen and spatial resolution. Because breast cancer screening involves exposing large populations of asymptomatic women to x rays in an effort to detect a disease with a low incidence (1.5–4.5 cases per 1,000 women per year) (11), radiation dose must be kept at an acceptably low level. However, the use of thin single screens and single-emulsion film required to achieve high resolution necessitates higher radiation doses than the double-screen, double-emulsion film technique used in other areas of general diagnostic

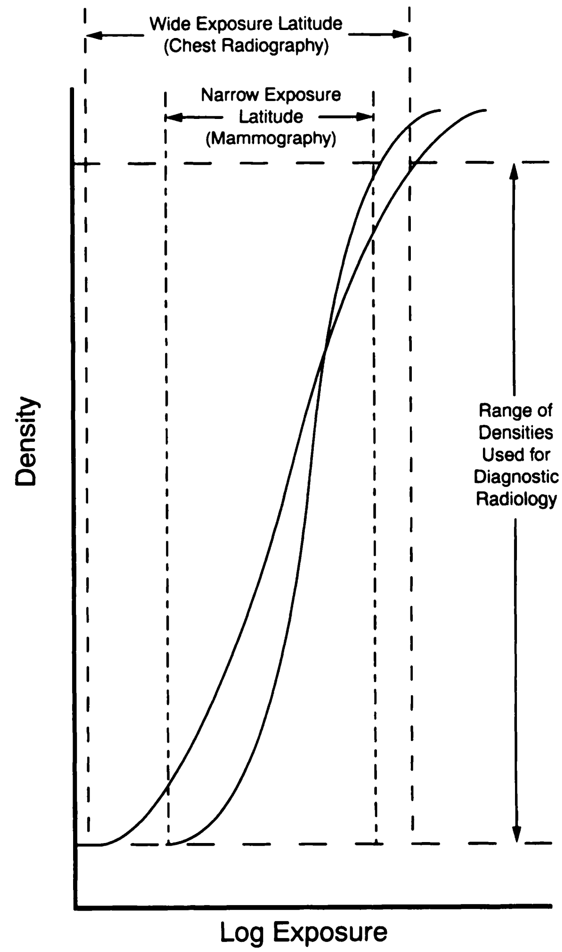
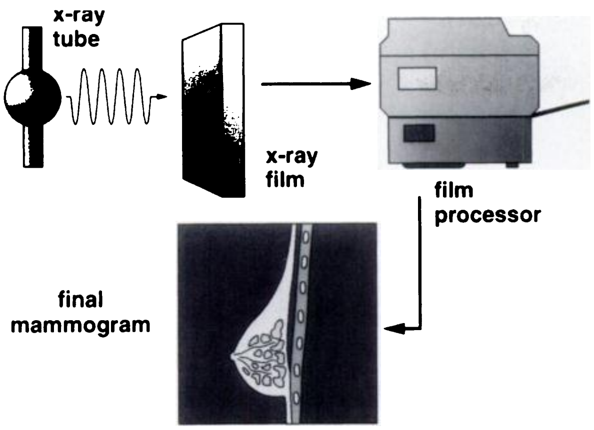


Figure 1. Graph shows typical H&D curves for screen-film mammography and chest radiography. The characteristic curve for mammography has a steeper slope, whereas the curve for chest radiography has a wider exposure latitude.

radiology. Yet another limitation of screen-film mammography is that overlapping dense tissues may obscure lesions despite high contrast levels and the use of breast compression (12).



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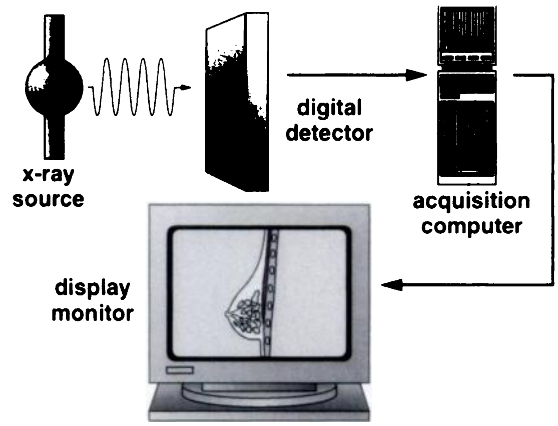


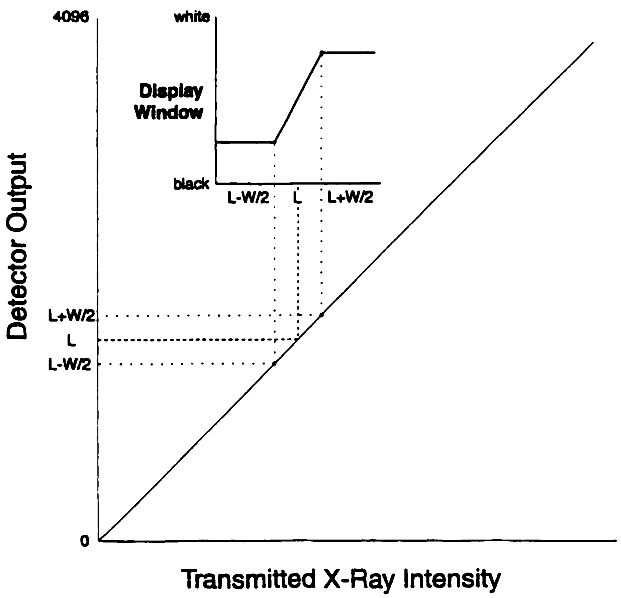
3.

Figures 2, 3. **(2)** Schematic illustrates the process for conventional screen-film mammography. **(3)** Schematic illustrates the process for digital mammography.

■ POTENTIAL ADVANTAGES OF DIGITAL MAMMOGRAPHY

For these and other reasons, the use of whole-breast digital mammography as an alternative to screen-film mammography is currently under investigation. With conventional screen-film mammography, x rays cause light emission from a phosphor screen, resulting in the formation of a latent film image. A single permanent film image is then produced with photographic processing (Fig 2). With digital mammography, the digital detector emits an electronic signal in response to x-ray exposure. This signal can be stored and processed on a computer (Fig 3). The image can then be displayed on a monitor or printed on film. In screen-film mammography, the film serves as a means of image acquisition, storage, and display; in digital mammography, these functions are performed by the separate components of a digital system, allowing multiple versions of the same image at different brightness and contrast settings, just as an image from a videotape may be altered by turning the knobs on a television set.

Digital mammography has numerous potential advantages over screen-film mammography. It has a wider dynamic range (a linear response of perhaps 1,000:1 compared with 40:1 for screen-film mammography). The digital mammogram has greater contrast resolution, especially in dense breast tissue. Because the electronic output is directly proportional to the transmitted x-ray intensity, digital mammography has a linear rather than a sigmoid response curve. Postexposure processing (windowing and leveling or other more sophisticated operations) may be performed to alter contrast and brightness, potentially allowing good contrast throughout all breast tissue (Figs 4, 5) (13).



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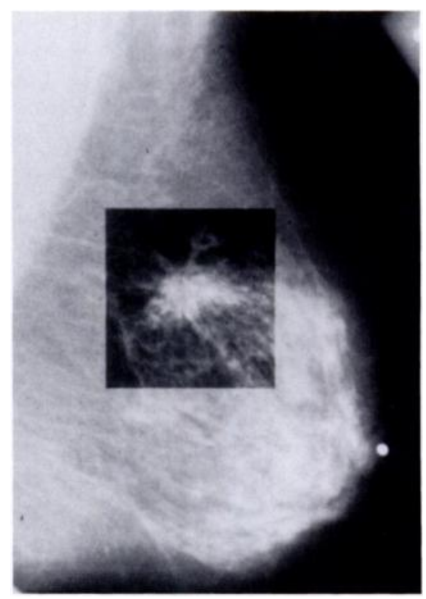
5a.



5b.

Figures 4, 5. **(4)** Graph shows the effect that manipulation of the level (*L*) and window (*W*) controls has on contrast enhancement in digital imaging. **(5a)** Conventional screen-film mammogram demonstrates a spiculated mass. **(5b)** Digital mammogram with image enhancement (inset) enables better visualization of the mass.

The digital image may have less noise because film granularity is eliminated. Radiation dose may be the same as or less than that used at screen-film mammography due to the increased detective quantum efficiency of the x-ray detector (14). On slot-scanning digital systems, radiation dose can be further reduced due to elimination of the grid (15). The fact that the diagnostic signs are identical for interpretation of screen-film mammography and digital mammography could facilitate the transition from one to the other. Digital mammography could potentially enable detection of breast cancer at an earlier stage, reduction of the number of patients recalled from screening for additional or repeat mammograms, and reduction of the number of false-positive biopsies.

■ POTENTIAL APPLICATIONS OF DIGITAL MAMMOGRAPHY

Digital mammography would also facilitate computer-aided detection and diagnosis (CAD) (16–18). Computer-aided detection may be defined as a second reading by a computer to in-

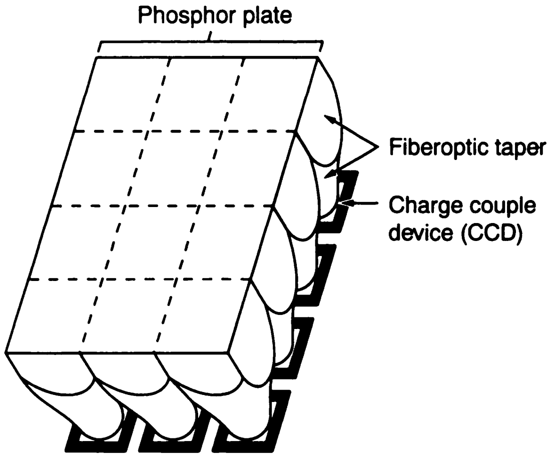


Figure 6. Illustration shows the detector assembly for a single-exposure multiple detector digital mammography system.

crease the cancer detection rate. Computer-aided diagnosis may also be used to provide a second opinion to estimate the likelihood of malignancy and reduce false-positive biopsy rates, thereby increasing diagnostic specificity. In the absence of digital mammography, CAD requires the cumbersome and time-consuming initial step of digitizing film images to provide suitable input for the computer. Digital mammography with CAD appears highly promising; however, further clinical studies are needed to evaluate the relative efficacy of different CAD methods.

Digital mammography also greatly facilitates the use of a picture archiving and communications system (PACS). A PACS reduces the need for storage space; increases retention of previous studies; and allows telemammography, the transmission of mammographic images from one location to another in digital format for expert consultation, off-site interpretation, monitoring of mammographic studies as they are performed, and comparison with images obtained at previous examinations stored at a remote site (18,19). A practical telemammography system requires that images be in digital form.

Although teleradiology systems are commonly used in many other areas of radiology, development of an adequate telemammography system is especially challenging due to the need for much higher resolution in mammography (20). There are trade-offs between the speed and cost of transmission, and image compression is performed to reduce cost and increase speed. However, the time required for image compression and decompression must also be considered. Image transmission may be either lossless (ie, no information is deleted from transmission) or lossy. Although the latter provides higher compression ratios, lossless transmission may be preferable for interpretation.

It is possible that some lesions may still be missed at digital mammography due to masking from overlying structures, particularly in dense breasts. Dual-energy subtraction imaging is one of the proposed solutions to this problem (21). By obtaining digital images with two substantially different x-ray spectra, the contrast of relevant structures would be preserved while unwanted masking contrast would be substantially removed. Digital tomosynthesis represents another approach to the masking problem. In this technique, multiple images are acquired as the x-ray tube moves in an arc above the stationary breast and detector (22). By manipulating the digital image, any plane in the breast can be clearly displayed.

■ DIGITAL MAMMOGRAPHY SYSTEMS

Different approaches to system design have been used by the three companies that are planning to market digital mammography systems in 1998. All of these systems employ a phosphor x-ray absorber whose signal is coupled to a photoelectric readout array.

In the large-area detector used in the digital mammography system developed by Bennett X-ray (Copiague, NY), a division of Thermotrex, an 18 × 24-cm cesium iodide scintillator is coupled to a mosaic of 12 charge-coupled devices (3 × 4 array) by 12 fiberoptic tapers (Fig 6) (23). A proprietary edge preparation and bonding technique allows a seamless image. Because the detector has no mechanical components, it can be retrofitted on the company's conventional mammography unit, which has a tilting capability. With use of a molybdenum-molybdenum or molybdenum-rhodium tube-filter combination, the unit operates at 23–28 kVp with an exposure time of 1 second or less. The detector is composed of 6,400 × 4,800 sensitive elements, each measuring approximately 40 μm on each side. Images are acquired at 14-bit precision and are viewed as hard copies printed with a high-resolution laser printer or on a video monitor with a 2.3 × 1.6-kpixel format.

A slot-scanning digital mammography unit has been developed by Fischer Imaging (Denver, Colo) (Fig 7). In this system, the image is acquired as an x-ray beam scans from side to side across the breast. The beam is confined to an approximately 24 × 1.4-cm slot that matches the format of the phosphor fiberoptic charge-coupled device detector array. The system produces less scattered radiation than other systems. However, mechanical synchronization between the detector array and the fan-shaped beam is necessary because both move. A time delay integration mechanism is used to avoid motion and stitching artifacts. Although the total scan time for the entire breast is 4 seconds, the time over any given slot is much less. However, because of the long exposure time, a tungsten tube is needed to accommodate heat loading considerations. Because of the scanning motion in the gantry, a specially designed unit is necessary.

The system developed by General Electric Medical Systems (Schenectady, NY) uses a detector consisting of a single solid-state flat panel in close contact with a cesium iodide phosphor (24). The flat panel is an amorphous silicon photodiode array consisting of 1,800 × 2,300 100-μm elements (Fig 8). The detector can be operated in an optional 50-μm oversampling mode in which the panel is automatically moved 50 μm along two orthogonal directions between each quarter of the exposure. The signals on the detector elements of this large matrix are read out by sending readout trigger pulses sequentially to each row of the detector and digitizing the electrical charge that appears on conductive lines connecting the elements down each column. The detector is retrofitted on the company's conventional mammographic unit.

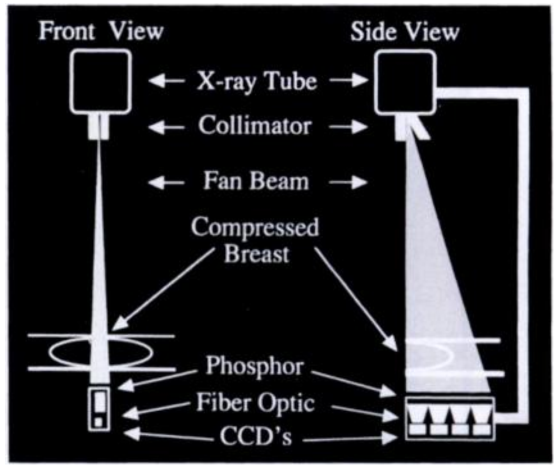


Figure 7. Schematic illustrates a slot-scanning digital mammography unit. *CCD* = charge-coupled device.

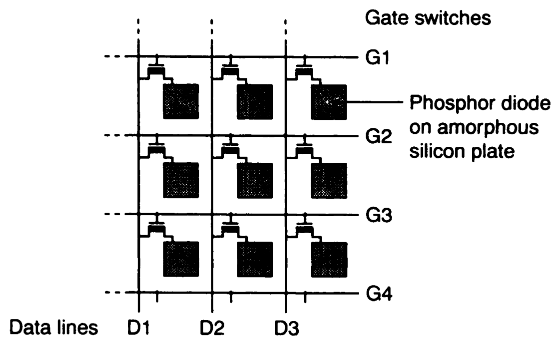
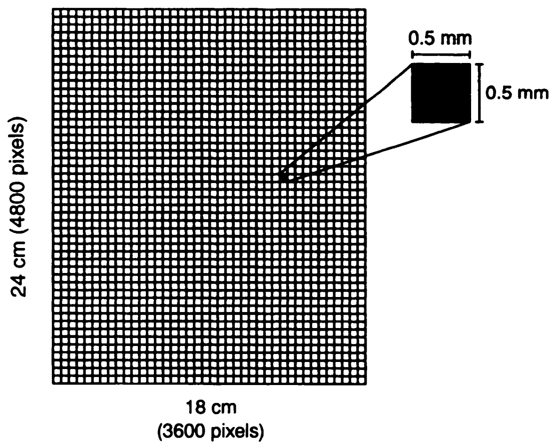


Figure 8. Diagram shows a flat panel single-exposure detector that makes use of amorphous silicon.

■ POTENTIAL LIMITATIONS OF DIGITAL DETECTORS

A digital image consists of a matrix of discrete picture elements, or pixels (Figs 9, 10). Pixel spacing places an upper limit on spatial resolution. However, spatial resolution also depends on other factors such as pixel size and light spread if a phosphor is used as the x-ray detector (14). The spatial resolution of current screen-film systems is often as high as 20 line pairs per millimeter (lp/mm) (equivalent to 25-μm pixels). However, screen-film mammography does not fully use the resolution capabilities of such systems because mammographic images have lower contrast and signal-to-noise ratio than are attainable with the high-contrast test objects used to measure spatial resolution. The higher contrast sensitivity and lower noise of digital mammography should allow for an acceptable resolution lower than 20 lp/mm.



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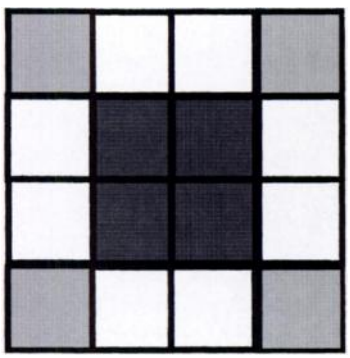
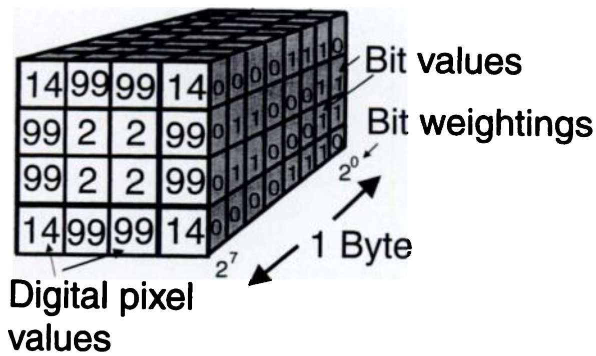


10a.



10b.

Figures 9, 10. **(9)** Illustration shows the pixel matrix for an 18 × 24-cm digital image. If each pixel measured 0.05 × 0.05 mm, there would be 1.728×10^6 (4,800 × 3,600) pixels. **(10)** Illustrations show the gray-scale levels in a digital image. The shade of gray for each pixel in the image **(a)** is determined by its digital pixel value, which is the sum of the product of bit value and bit weighting for each bit in the pixel (eg, $[1 \times 2^1] + [1 \times 2^2] + [1 \times 2^3] = 2 + 4 + 8 = 14$) **(b)**.

However, it is likely that characterization of tiny objects such as microcalcifications may require at least 10 lp/mm (50-µm pixels) (14).

■ LIMITATIONS OF CURRENT MONITORS

Display of an image of the entire breast at a resolution of 10 lp/mm requires a 4 × 5-kpixel monitor. Although such monitors have been developed for military and selected industrial applications, they are prohibitively costly for widespread medical use. Furthermore, although anticipated technologic advances may allow cost reduction and increased availability of such monitors, currently available monitors have a resolution of only 5 lp/mm with a 2.0 × 2.5-kpixel field of 100-µm pixels. Because these monitors have less resolution than whole-breast digital detectors, digital images may need to be displayed on laser-printed film as an interim solution, although the use of such film adds time, cost, and equipment to the digital process.

Currently, display of digital images on a monitor may vitiate some of the other advantages of digital detection. The monitor itself may add noise to the image. The light output from a monitor is considerably less than that from a high-intensity view box, although this may not be a problem if appropriate display strategies are used. The contrast resolution of monitors is limited to 8–9 bits; digital detectors have a depth of 12–14 bits, enabling display of many more gray-scale levels (Fig 10).

■ DESIGN OF A USER-FRIENDLY WORKSTATION

The full potential benefit of digital mammography is unlikely to be achieved until "soft-copy" images can be interpreted on a workstation. The requirements for a user-friendly workstation include the capability for viewing and comparing eight images from current and previous examinations and for optimizing the gray-scale presentation conveniently. Such a workstation will have to allow images to be selected, moved, and matched and lesions marked with the same speed and ease that radiologists take for granted when they review images on a view box and circle lesions with a grease pencil.

■ COST CONSIDERATIONS

Although digital mammography may reduce some costs by eliminating film and film processing chemicals, decreasing film storage space, and reducing film library staff, immediate cost savings are unlikely. Conversion to digital mammography will entail substantial capital equipment costs because digital mammography systems will cost \$250,000–\$500,000 each, an amount four to five times the cost of conventional mammography units. Laser printers and 4 × 5-kpixel monitors will entail additional costs. Determination of the relative cost-effectiveness of digital mammography versus screen-film mammography will require outcome studies to document improved detection of earlier stage cancers and cancers in dense breasts, decreased false-positive biopsy rates, or decreased work-up rates for digital mammography. Current reimbursement rates for screening mammography (about \$50 per case) are among the lowest for all imaging procedures and may not be sufficient to support increased expenditures for digital mammography.

■ FUTURE DEVELOPMENTS

Several possible technologic advances could result in improved image quality, better clinical efficacy, lower radiation dose, and decreased costs for digital mammography systems. Development of reasonably priced, user-friendly workstations that can handle the required number and size of digital mammograms and allow rapid image navigation could allow soft-copy interpretation. Direct conversion of x rays to electronic signal by means of thin film transistors made with amorphous silicon and covered with selenium or other solid-state materials would eliminate intermediate steps and allow lower radiation dose and higher resolution. There should also be development of image display algorithms for breasts of different composition or for different areas of the same breast having different composition. Automatic, near optimal display of images with override capability for the radiologist is highly desirable. Requiring either technologists or radiologists to spend excessive amounts of time performing windowing and leveling for each image to optimize image quality is not a practical alternative.

In the future, higher-resolution flat panel displays, possibly based on liquid crystal display technology, may produce images with higher resolution, less noise, and more gray-scale levels than can be produced with current monitors. Alternatively, improved strategies of image navigation that will make possible the transition between a low-resolution gestalt (overview) and a higher-resolution depiction of areas causing suspicion may allow adequate display on existing hardware.

■ CONCLUSIONS

Although whole-breast digital mammography offers potential advantages in the detection and diagnosis of breast cancer and in image storage and transmission, major technical challenges remain, and further clinical studies are needed to determine the actual clinical value of this modality. It is necessary to determine whether and how often digital mammography can enable detection of more cancers and earlier stage cancers than screen-film mammography and the impact of digital mammography on the need for additional mammograms and on false-positive biopsy rates. It is also important to determine how the additive value of CAD will vary according to the radiologist's skill and expertise and the quality of the images. The relative value of different CAD algorithms and their effect on sensitivity and specificity as well as the effects of CAD on interpretation time and false-positive rates need to be studied. Moreover, the medicolegal status of CAD is uncertain at present and needs to be clarified.

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