Eur Radiol (2006) 16: 38–44
DOI 10.1007/s00330-005-2848-0

BREAST

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Digital mammography: current state and future aspects

Received: 15 February 2005
Revised: 9 June 2005
Accepted: 21 June 2005
Published online: 20 August 2005

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Abstract The introduction of digital technique in mammography has been the last step in completing the process of digitalization in diagnostic imaging. Meanwhile, some different digital techniques as well as a couple of different digital mammography systems were developed and have already been available for some years. In this review article, the relevant data of key studies are reported, the current status is defined, and perspectives of digital mammography are described.

Keywords Digital mammography · Breast imaging · Imaging modalities

Introduction

X-ray examination of the breast (mammography) has been the last area of roentgenology to make the transition from analog to digital imaging. It was assumed that digital mammography would require similar spatial resolution as that of the high-resolution screen-film systems used in conventional mammography, and that digital techniques would be limited by the pixel size of the digital detector in the detection of small structures such as microcalcifications. However, it has been shown that other characteristics of digital systems more than compensate for the lower spatial resolution leading to an enhanced detection of microcalcifications. In women with an inhomogeneous or extremely dense parenchymal pattern (type III and IV of the American College of Radiology, ACR) the detection of breast cancer is difficult due to the similar X-ray absorption of carcinoma and the surrounding normal dense breast tissue. In this context, digital detectors may offer improved detection of carcinomas thanks to a higher efficiency of absorption of X-ray photons, a linear response over a wide range of incident radiation levels and a low system noise. The goal of this paper is to review the key studies, assess the current status and propose a perspective of digital mammography.

History

The first attempts to make a digital radiography system started in the 80s with the introduction of digital subtraction imaging and computed radiography (CR) with photostimulable storage phosphors. Systems using charge-coupled devices (CCD) were the next step for stereotactic preoperative localizations and percutaneous biopsies. However, CCD chips could not be used for full-field breast imaging due to the limitations of the field of view (chip size). The CR systems derived from general radiography were also used for full-field breast imaging. These systems did not gain wide acceptance due to limitations in detective quantum efficiency (DQE) and low spatial resolution [1–3]. Other working groups combined digital storage phosphor plates with the direct magnification technique and an X-ray tube with a very small focal spot [4]. The results were encouraging and magnification mammography with this technique was demonstrated to be superior to magnified conventional mammography in the detection of microcalcifications. This system also failed to get wide acceptance and further development of this combination ended in the late 90s. Today a variety of detectors are used with digital mammography systems: high-resolution photostimulable storage phosphors, cesium Iodide (CsI) scintillators combined with an amorphous silicon (a-Si) photodiode matrix or CCD arrays, a photoconductive layer of amorphous selenium (a-Se) with thin-film transistor arrays, and crystalline silicon used as a photon counter. A significant number of publications exists on the CsI/a-Si system—Senographe 2000D (GE Medical Systems, Waukesha, WI, USA) and only a few on other digital systems [5–7]. Consequently, most of the results mentioned in this article were demonstrated for Senographe 2000D.

Digital mammography systems

From a user's perspective, two different technologies have been differentiated in digital mammography: off-line systems with a cassette-based removable detector using an external reading device to generate the digital image; and on-line systems, where the detector is integrated into the digital mammography unit and the digital image data are directly read by the system in quasi real time. For off-line systems, commonly designated as CR systems, a high-resolution storage phosphor plate (FCR 5000MA, FujiFilm, Tokyo, Japan; CR 850, Kodak, Rochester, NY, USA; Regius 190, Konica-Minolta, Tokyo, Japan; ADC Compact plus, Agfa, Mortsel, Belgium) is usually used in combination with a conventional mammography unit. The first online system for digital mammography which received FDA approval for general mammography utilizes a CsI scintillator and an a-Si photodiode/transistor array. This system will be referred to as CsI/a-Si and is manufactured by G E Medical Systems (Waukesha, WI, USA). The CsI phosphor layer is used to convert X-ray photons to light photons, which are then converted into an electrical signal by the photodiode array and read out by transistor array. The second approved system is manufactured by Fischer Imaging (Denver, CO, USA) and is based on a CsI phosphor and a line detector composed of four CCD sensors. This system utilizes a fan beam of radiation that scans across the breast in approximately 6 s. The fan beam is effective in removing scattered radiation and therefore this system does not require an anti-scatter grid. This system will be referred to as CsI/CCD. Another detector based on a photoconductive layer of a-Se with a thin-film transistor array, has also been approved by the FDA for clinical use. This system will be referred to as a-Se/a-Si and is manufactured by Hologic (Bedford, MA, USA). Selenium allows for the direct conversion from absorbed X-ray into electronic charges which are read out by arrays of thin-film transistors. Se-based systems are in manufacturing process by Instrumentarium Imaging (Tuusula, Finland—currently with GE Medical Systems), Siemens (Erlangen, Germany), IMS (Bologna, Italy) and others. Some of these systems are currently under evaluation by the FDA. All detectors mentioned above accumulate the signal from all of the X-ray photons that fall on a detector pixel to create an image. The individual counting of each interacting X-ray photon, an X-ray photon produces exactly one count regardless of its energy, is another technology for building a direct converting digital detector. The MicroDose system (Sectra, Linkoeping, Sweden) is based on this technology.

Although other systems are under development, Table 1 includes either well-proven or currently implemented large-scale clinical testing systems representing four different technological approaches to digital mammography.

Table 1 compares a number of detector features. Pixel size varies from 50 to 100 μm with the CsI/CCD detector having the smallest pixel size. Image area also varies with the a-Se/a-Si and CR systems having the largest imaging areas. Larger imaging areas may allow a higher fraction of women to be imaged with one exposure than those systems with a smaller imaging area. The amount of data per image also varies by a factor of more than 6. The large amounts of data may be a consideration for image archive and display. Images with large matrix sizes cannot be displayed at full resolution with current soft copy displays since these displays are limited to approximately 2,000×2,500 pixels. Three of the systems use a grid while the CsI/CCD system uses a fan beam and, therefore, does not require a grid. Anode material may also be important in determining the radiation quality and image quality. The systems have anode materials of Molybdenum, Rhodium, or Tungsten. The CsI/a-Si, a-Se/a-Si and CR systems have the potential of using geometric magnification while the scanning CsI/CCD system has a high-resolution mode but does not have geometric magnification. The as low as reasonably achievable (ALARA) principle on dose administered to the patient urges the use of an automated exposure control (AEC) system to ensure the optimal exposure of the image receptor compensating for breast thickness and composition. The use of reference tables based only on the measured thickness of the compressed breast is insufficient and increases the patient dose. While the flat-panel systems operate with sophisticated systems for automatic optimization of the radiation parameters, the scanning system determines exposure based only on breast thickness. The following review will discuss research studies pertaining to various aspects of digital mammography.

Image quality

The image quality obtained with a CsI/a-Si digital mammography system (Senographe 2000D, GE Medical Systems), a CR system (Fuji FCR 5000MA with Siemens Mammomat 3000N) and a conventional film-screen mammography (SFM) system (Siemens Mammomat 3000N+

Table 1 Digital mammography system comparison

Manufacturer	General Electric	Fischer	Lorad	Fuji
FDA approval	Yes (2000)	Yes (2001)	Yes (2002)	Yes (2004)
Detector type	CsI/a-Si, flat panel	CsI/CCD, slot fan beam	a-Se/a-Si, flat panel	CR, storage phosphor cassette
Pixel size (μm)	100	54 (27[a])	70	50
Image area (cm)	19.2×23	21×29	25×29	18×24 24×30
Detector area (cm)	19.2×23	21×1	25×29	18×24 24×30
Matrix Size	1920×2304	4096×5625	3584×4096	3540×4740 4728×5928
Bits	14	12	14	10
Image data size (Mbytes)	8.85	46.1	29.4	33.6 56.1
Exam data size (Mbytes)	35	184	117	134 224
Scatter removal method	Grid	Slot	Grid	[b]
Anode target material	Molybdenum, Rhodium	Tungsten	Molybdenum	[b]
Filter material	Molybdenum, Rhodium	Aluminum	Molybdenum, Rhodium	[b]
kVp range	22–49	20–45	22–39	[b]
Magnification	Geometric	Electronic	Geometric	[b]
Parameters controlled by AEC	Target, Filter, mAs, kVp	–	Optionally: mAs; mAs and kVp; Filter, mAs and kVp	[b]

[a]High-resolution mode has a 27-μm pixel size

[b]This system is a detector only and may be used with many different mammography machines. The geometry, automatic exposure control, target material, filter material and kVp range will be determined by the mammography system used with the detector

Fuji UM Mammo Fine UM-MA) was compared using the Wisconsin Mammographic Random Phantom Model 152A [8]. The CsI/a-Si digital mammography system was superior to the other imaging modalities with a mean number of depicted details of 40.4 in comparison to CR mammography (mean 38.6) and SFM (mean 38.2). The entrance surface air-kerma for SFM, CR and CsI/a-Si systems were 9.64, 7.60, and 7.02 mGy, respectively [8]. Obenauer et al. compared image quality of CsI/a-Si and SFM in 55 patients with cyto- or histologically proven lesions [9]. In addition, 75 CsI/a-Si digital mammograms of normal patients (no lesions) and their previous SFM examinations (not older than 18 months) were also included in the study. Three readers evaluated image quality (contrast) which was rated as good in 76%, satisfactory in 20% and poor in 4% of cases for SFM compared to 99, 0 and 1% for the Cs/aSi system.

Take home point: Image quality of digital mammography seems at least equivalent to SFM.

Spatial resolution

For some years, image quality of conventional and digital mammography systems has been assessed with a lead bar pattern determining the limiting spatial resolution quoted by the number of yet visible line pairs per mm. Full-field digital mammography (FFDM) systems have a lower-limit spatial resolution than SFM and were therefore considered inferior. However, relevant features in the breast may have low or medium contrast and their detection is not limited by spatial resolution at high contrast but rather by contrast resolution [9]. Therefore, spatial resolution is not the only relevant criterion for defining the quality of a digital mammography system. Frequently the detective quantum efficiency (DQE) is considered to be the best parameter to describe the imaging performance of an X-ray detector. The DQE describes the transfer of signal to noise ratio (SNR) as a function of spatial frequency when recording an X-ray image. The DQE gives the efficiency with which the detector uses the available quanta. However, this parameter is not well suited to qualify the performance of a whole mammographic system. Furthermore, because there is no international standard which defines the measuring condition or the measuring method of the DQE for detector systems used in digital mammography, values from different working groups in many cases are incommensurable. Since the determination of DQE requires a formidable measuring expenditure, usually the imaging performance is evaluated in terms of threshold contrast visibility.

Take home point: It is not spatial resolution but rather contrast resolution in terms of threshold contrast visibility that is the best choice in characterizing and comparing the imaging performance of mammographic systems. Beware of comparing DQE values from different sources.

Dose aspects

The average glandular dose delivered by digital systems has been evaluated from phantom studies as well as patient examinations. However, it is well known that the measurement of the radiation dose delivered to an acrylic or another standardized phantom does not provide the complete information about patient dose in mammography.

Phantom studies demonstrated the dose reduction potential of full-field digital mammography systems. A well-known contrast-detail phantom which combines objects of variable contrast and size is the CD-MAM (e.g., CDMAM 3.4, N University Medical Center, Nijmegen, The Netherlands). A 25% reduction of the radiation dose led to equivalent results in the detection of microcalcifications on CDMAM phantom compared to film-screen mammography. On the other hand, the detection threshold of the digital system increased in a range of 10–25% when an equivalent dose level was used [10, 11].

Hermann et al. presented a clinical study based on data from 1,116 mammograms of 591 women who were examined with a Cs/aSi FFDM unit (Senographe 2000D, GEMS). The mean average glandular dose was 1.51 mGy for a single view. The authors stated that the digital system needs about 25% less dose in comparison with conventional screen-film mammography [12]. Gennaro et al. evaluated four different CsI/aSi systems in routine clinical use and reported a mean average glandular dose ranging from 1.25 to 1.89 mGy depending on the breast glandular composition [13].

The possibility of using higher energy exposures by various anode track-filter combinations may result in additional patient dose reduction with no loss in clinical performance. As an example, a Mo/Rh anode-filter combination plus a tube voltage of 31 kVp (25% dose reduction) and a Rh/Rh combination at 32 kVp (37% dose reduction) did not show significant differences in the ROC analysis for the detection of microcalcifications in comparison to the established Mo/Mo combination at 28 kVp in an anthropomorphic phantom [14]. Due to higher DQE and therefore lower noise, digital systems can compensate for the relative loss of contrast via appropriate window settings [15].

Take home point: The potential of dose reduction with FFDM as compared to SFM ranges from 30 to 50%.

Soft-copy reading (monitor reading)

Full-field digital mammography allows direct image analysis from high-resolution monitors (soft-copy reading). No significant difference between soft-copy and hard-copy reading was seen in the presentation of masses, microcalcifications, and parenchymal tissue [16, 17]. In a large population-based mammography screening study of 3,683 women, Skaane et al. did not find any statistically significant difference in cancer detection rate between SFM and FFDM with soft-copy reading [18]. Considering the speed and accuracy of soft-copy vs. printed-film display, Pisano et al. did not find statistically significant differences in speed of interpretation, receiver operating characteristic curve A(z), sensitivity, and specificity. However, A(z) and sensitivity were slightly better for interpretation with printed film and specificity was slightly superior with soft-copy reading [19].

Take home point: Soft-copy and hard-copy reading are equivalent in the interpretation of digital mammograms.

Detection and characterization of masses

Based on a phantom study, the 2000D flat panel system was shown to be superior to conventional screen-film mammography using an equivalent radiation dose. The correct observation ratio was 0.95 with FFDM compared to 0.82 with SFM. Regarding equal detection rates regarding both techniques, FFDM allowed a dose reduction of 40% compared to the dose of the conventional images [20].

Breast lesion characterization was checked by Kuzmiak et al. who evaluated 79 breast surgical specimens (37 masses) using SFM and FFDM. Their results demonstrate that digital FFDM and SFM have similar diagnostic accuracy [21].

Take home point: FFDM seems at least equivalent to SFM in the detection and characterization of masses.

Detection and characterization of microcalcifications

FFDM has been shown to be superior to conventional mammography in the detection and characterization of microcalcifications in different phantom and in vivo studies. Using an anthropomorphic breast phantom with simulated microcalcifications, the digital technique was shown to be equivalent for contact mammograms (A_z 0.68 vs. A_z 0.63 in ROC analysis) and superior for spot views (A_z 0.79 vs. 0.70) in comparison to conventional mammography [22]. Regarding in vivo specimens, FFDM was also superior to SFM in the presentation and detection of microcalcifications based on the number of counted particles [23].

In the detection and characterization of microcalcifications in 57 patients, Fischer et al. performed a comparison of FFDM and SFM images obtained from the same woman. Sensitivity and specificity for FFDM vs. SFM were 95.2 vs. 91.9% and 41.4 vs. 39.3%, respectively. Moreover, FFDM demonstrated a higher diagnostic accuracy (deviation in BI-RADS steps vs. truth was 0.86 for FFDM and 0.93 for SFM) [24].

Take home point: FFDM is at least equivalent to SFM in the detection and characterization of microcalcifications.

Image processing

Digital data offers the potential for postprocessing including grey scale inversion, modification of the window width and window level (contrast and brightness), zooming, magnification function, contour enhancing and image subtraction. In a comparison between direct magnification mammography and postprocessing zooming of contact mammography in patients with microcalcifications, there was a mild advantage of direct technique considering the sensitivity (magnification vs. zooming; 97.5 vs. 96.3%). Zooming was superior to direct magnification for specificity (40 vs. 34.3%), positive predictive value (49.8 vs. 47.5%) and accuracy (61.2 vs. 58.1%) [25]. In conclusion, digital zooming from contact mammography may reduce the need of additional magnification images and decrease the total examination dose.

Mammography requires specific image processing for different reading purposes (screening vs. diagnosis) or different lesions (calcifications vs. masses) [27]. Evaluation of three different digital mammography systems showed that the radiologist's interpretation accuracy depends on lesion type (calcification vs. mass) in women with mammographically dense breasts. However, interpretation accuracy was not influenced by the image-postprocessing method in this study [27].

Take home point: Postprocessing tools are helpful in the evaluation of digital mammography images. However, an increase of diagnostic reliability has not yet been proven.

Computer-aided detection (CAD)

It has been substantiated that computer-assisted detection improves the radiologist's sensitivity. CAD systems are natural tools for the interpretation of digital mammography images. Three major CAD systems are currently being used with FFDM systems including (1) Image Checker (R2 Technology, Sunnyvale, CA, USA), (2) Look (CADx Systems, Beavercreek, OH, USA), and (3) the Mammex TR (Scannis, Foster City, CA, USA). CAD using directly acquired digital data and CAD based on digitized film provided equivalent results. Moreover, CAD applied to FFDM provides an optimized workflow. The sensitivity of CAD for detection of microcalcifications is acceptable. The sensitivity of CAD systems is lower for masses than for microcalcifications [28]. Other authors have demonstrated the value of CAD within a large screening population of more than 12,000 women. They found an increase in detection of early-stage malignancies without a significant increase in the recall rate [29]. In conclusion, the development of CAD-systems in the past years is promising and improvements in the future are expected.

Take home point: CAD is a helpful tool in the detection of early stage malignancies.

Digital mammography in screening population

In a population-based screening program (Oslo I), FFDM was compared to SFM in 3,683 women aged 50–69 years [18]. The difference in cancer detection rates was not significant and side-by-side cancer conspicuity was equal with both modalities; however, the recall rate was higher for FFDM. In another paired screening study involving 6,736 patients [30], no statistical difference was found in cancer detection between SFM and FFDM, but the recall rate was shown to be significantly lower with FFDM. These initial studies were carried out with prototype FFDM units, suboptimal soft-copy reading conditions or readers with limited soft-copy experience. In a more recent randomized study involving more than 25,000 women and using production-level FFDM systems and optimized soft-copy reading, the cancer detection rate in the age group of 50–69 years was found superior for FFDM (very close to statistical significance), and similar for the age group 45–49 years. Recall rate was higher for FFDM but at a low level [31]. As screening programs involve large populations of healthy patients, digital mammography should offer considerable benefits in terms of radiation dose, image quality, data transfer (for double reading, telemammography) and data archiving.

Take home point: FFDM should become the method of choice for breast-cancer screening. Advantages would be, i.e., dose reduction, CAD analysis, telemammography, and digital data archiving.

Digital mammography in high-risk women

Patients with genetic breast cancer susceptibility have an increased vulnerability to ionized radiation. Therefore, extended and short-interval X-ray mammography is not recommended for these women. MR imaging of the breast has been demonstrated to be superior in this high-risk group [32]. A combination of one-view digital mammography (mediolateral oblique) to depict microcalcifications at a lower dose and contrast-enhanced MR mammography (so-called "Optipack" concept) to demonstrate invasive breast cancer, seems to be an appropriate protocol in the evaluation and follow-up of women with hereditary risk of breast cancer. This combination may provide the highest sensitivity in the detection of breast cancer (intraductal as well as invasive) with the lowest radiation dose.

Take home point: FFDM may be a relevant part in a combined protocol using one-view low-dose mammography plus contrast-enhanced MR in high-risk women.

Perspectives and future aspects

Rapid acquisition of several digital images at different X-ray beam angulations should make three-dimensional mammography a reality in the near future. The process by which several images can be combined and presented as tomographic planes or three-dimensional reconstruction is called tomosynthesis. First specimen tomosynthesis images have been presented in 1997 by Niklason et al. [33]. This technique may improve the specificity of mammography due to better visualization of lesion shape and border, especially in women with radiographically dense breasts.

Stevens and co-workers presented preliminary results of circular tomosynthesis in breast cancer using a phantom and mastectomy specimen. As expected, it was shown that breast tissue visualization is improved by the ability to produce sectional images that blur overlaying structures and yield three-dimensional information about calcification clusters [34].

Dual-energy subtraction is another technique which may become practical with digital mammography. Two different exposures are taken, one at low energy (i.e., 25–30 kVp) and one at higher energy (i.e., 40–50 kVp). X-ray absorption depends on exposure energy and breast tissues are represented differently in the two images. Various processing algorithms as a weighted subtraction of high- and low-energy images can be performed to obtain additional information about breast-tissue composition. Data suggest benefits for imaging calcium which may enhance microcalcifications, especially in dense parenchyma [35].

Breast cancer detection may also be improved by looking for increased vascularity and blood flow (angiogenesis). Just as increased uptake of contrast material in MRI may display breast cancer, digital mammography should also be able to depict differences in the enhancement of malignant lesions. Preliminary results using contrast-enhanced digital mammography are reported by M. Yaffe and coworkers. They found enhancement in 10/12 histologically proven carcinomas, while one DCIS and one IDC were missed. No uptake of iodine contrast material was observed in 7 of 12 cases with suspicious lesions in standard mammography which were proven to be benign [36]. Moreover, newly designed contrast media could optimize contrast-enhanced techniques in digital mammography [37].

Another development describes the potential combination of a digital mammography system with an integrated breast ultrasound to provide X-ray and ultrasound image registration. Clinical studies are needed to evaluate workflow, sensitivity, and specificity of this technique.

Take home: The value of FFDM could be increased with new techniques like tomosynthesis, dual-energy imaging, contrast enhancement, and ultrasound image fusion.

Limitations

The main disadvantage of digital mammography is the initial cost of the system and the discrepancy between this cost and the current rate of reimbursement. Other aspects like limited detector size or the lack of an automatic exposure control of some systems should be solved in time.

Final statement

Digital full-field mammography should become the method of choice in the detection and characterization of breast cancer.

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