Mammography Equipment

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Ioannis Sechopoulos and Cláudia Sá dos Reis

Learning Objectives

- Understand the basic function of the different components of a mammography system.
- Describe the similarities and differences between the equipment used for the different x-ray-based breast imaging modalities.
- Describe the basic concepts of image processing and display after image acquisition.

Introduction

Being one of the most technically demanding examinations in radiography, mammography requires X-ray technology designed specifically for the task. The pathology to be imaged ranges from small, 20–100 μm, high density calcifications to ill-defined low contrast masses. These must be imaged against a background of mixed densities. This makes demonstrating pathology challenging. Because of its use in asymptomatic screening, mammography must also employ as low a radiation dose as possible while still yielding an adequate image for the task at hand [1, 2].

In the past three decades, revolutionary technological developments in breast imaging have taken place [1, 2]. The main goal pursued by the mammography equipment industry has been to develop practical, inexpensive, and harmless equipment which is both appealing and effective in identifying, localising, and characterising abnormal tissues and signs of pathology within the breast [3–5]. Currently available technologies for breast imaging are used to identify structural or morphological differences in tumours, such as calcifications, soft tissue masses, asymmetry and architectural distortion. Some of the more recently developed techniques, especially with the use of intravenous contrast agents, can provide information about the biological or functional differences between tumours and normal tissues.

Mammography is based on the differential attenuation of X-ray photons in the breast tissues and this process is optimised when low-energy photons are used [6, 7]. The varying composition and densities of the adipose and fibroglandular tissues produce singular contrasts represented as

I. Sechopoulos (✉)
Department of Medical Imaging, Radboud University Medical Center, Nijmegen, The Netherlands

Technical Medical Centre, University of Twente, Enschede, The Netherlands

Dutch Expert Centre for Screening (LRCB), Nijmegen, The Netherlands
e-mail: ioannis.sechopoulos@radboudumc.nl

C. S. dos Reis
Escola Superior de Tecnologia da Saúde de Lisboa, Lisbon School of Health Technology, Lisbon, Portugal
e-mail: claudia.reis@estesl.ipl.pt

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dark and bright areas in the mammography image. However, the composition and density of fibroglandular tissue and carcinoma are similar and therefore, for the most part, non-calcified lesions are distinguished by their morphology, not their contrast.

The need for lower energy X-ray spectra to enhance contrast, for high spatial resolution to aid in the visibility and characterisation of calcifications, and for breast compression to reduce tissue overlap (among many other reasons), promotes the refinement of dedicated X-ray equipment for mammography. These systems count with specialised X-ray tubes, detectors, and overall acquisition geometries [6].

Technological advances over the last several decades have greatly improved the diagnostic performance of mammography. However, probably the most important milestone was the introduction of digital mammography systems in the 1990s. The possibility of advanced acquisitions and post-processing of the images opened the door to new imaging techniques that have great expanded the possibilities of X-ray breast imaging. Over the last two decades, three new X-ray modalities have been developed aimed at addressing the two most important limitations of standard mammography, its two-dimensional nature and its anatomical-imaging only nature.

Digital breast tomosynthesis (DBT) [8–10] and dedicated breast computed tomography (BCT) [11–13] have finally introduced tomography, to varying degrees, to X-ray breast imaging. The former, although only a pseudo-three-dimensional modality, shares the same platform as standard mammography, and therefore results in equivalent installation requirements and very similar workflow and interpretation. This has established DBT as the new standard in X-ray breast imaging, in many countries practically replacing standard mammography completely [14]. BCT involves a completely new system geometry and acquisition setup. However, BCT imaging requires no breast compression, making it attractive for the patients, and is fully three-dimensional, promising to further enhance clinical performance. Therefore, interest in BCT is high, and its best placement in the clinical realm remains to be ascertained.

While still two-dimensional, contrast-enhanced spectral mammography (CESM) results in functional images in which the blood flow and uptake in (malignant) lesions are depicted [15–17]. This capability has made CESM be of great interest, being a potential alternative to breast MRI while being considerably more affordable and with fewer contraindications.

The Mammographic X-Ray Unit

Mammography is performed using dedicated equipment usually with a "C" shaped arm aimed at facilitating breast positioning. The C arm can be adjusted in height and angular orientation to adjust the compression paddle and the breast support to the client standing or sitting position. The X-ray tube and digital receptor table assembly are mounted in opposition: the X-ray tube for the generation of the photon beam along with a face protector on the top head, and a compression paddle, the image receptor system on the lower arm (Fig. 18.1).

The stages for production of mammography images are acquisition, processing, display, and post processing for interpretation and storage. In digital mammography, each step is performed by an individual system that can be independently assessed and optimised. The image acquisition system is composed of an X-ray tube, breast compression paddle, and image receptor system [7]. The distance from the X-ray focus to the breast support platform is commonly around 60 cm. A moving anti-scatter grid is normally used, which is situated just behind the low-attenuation (often carbon-fibre) tabletop and in front of the image receptor. Some designs work without an anti-scatter grid and make a software correction for the large-scale effects of scattered radiation in the image [18, 19].

Due to the requirements for very high spatial resolution, X-ray focal spot sizes must be small. Focal spots of approximately 0.3 × 0.3 mm are used for conventional mammography, with a size

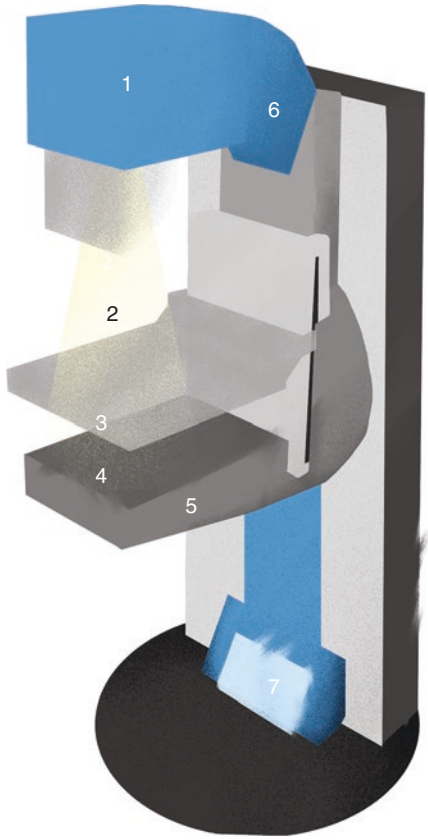

Fig. 18.1 Integrated direct digital mammography system. 1 X-ray tube, 2 X-ray beam, 3 compression paddle, 4 breast support, 5 detector, 6 C-arm, 7 monitor for angle, breast thickness, and compression force

of 0.15 × 0.15 mm selectable for magnified views, where the breast is raised away from the image receptor on a special magnification table to produce a geometrically magnified view [7].

The X-ray tube is positioned within the unit so that the anode heel effect is employed to reduce X-ray intensity towards the nipple side of the field where the breast will be thinner. Heavy reliance is placed on the automatic exposure control system of modern mammography units. These systems can sense the thickness and composition of the compressed breast and then automatically select the tube potential, target and filter combination, and tube current-exposure time product required to give the optimal imaging exposure within the constraints of patient dose limitations [7].

The Mammographic X-Ray Spectrum

The X-ray spectrum from a conventional tungsten target, glass encapsulated, aluminium filtered X-ray tube is not necessarily optimal for mammography. The optimal photon energy considering subject contrast between the different breast tissues and dose is around 20 keV, which is much lower than that used in normal radiography [7]. Increasing photon energy will reduce contrast and reducing photon energy will lead to inadequate penetration of the breast and a large increase in patient dose, so the X-ray spectrum is critical. A range of mammographic spectra are used for digital mammography.

The X-ray tube target may well be switchable (depending on the design) between molybdenum and rhodium, or consist only of tungsten, which has now become the most common X-ray tube target in mammography systems. Due to the low-energy spectra, to maximize tube output, the tube has a low attenuation beryllium output window. The beam is then filtered with either molybdenum, rhodium, silver, or aluminium filters, with other filter materials, like copper, titanium, or tin, being used or investigated for use in CESM [8, 9]. The X-ray tube is operated, in general, at a voltage in the range of 25–35 kV, although somewhat higher voltages may be used for DBT, while, of course, CESM requires the use of voltages in the range of 45–49 kV for the acquisition of the second, high-energy projection.

Figure 18.2 shows the spectrum of a rhodium target, rhodium filtered beam at a tube voltage of 30 kV. Rhodium has characteristic X-ray peaks at 20.2 and 22.7 keV, which contribute strongly to the limited range spectrum. Rhodium is again used as the filter because, due to the K-edge absorption, it strongly attenuates energies just above its own K-characteristic peaks as well as attenuating lower energies. The end result is a spectrum with most photons lying in a narrow band of energies.

Although molybdenum and rhodium targets are no longer the most common X-ray sources used in mammography, these are still in use and Fig. 18.2 is a good example of how the right com-

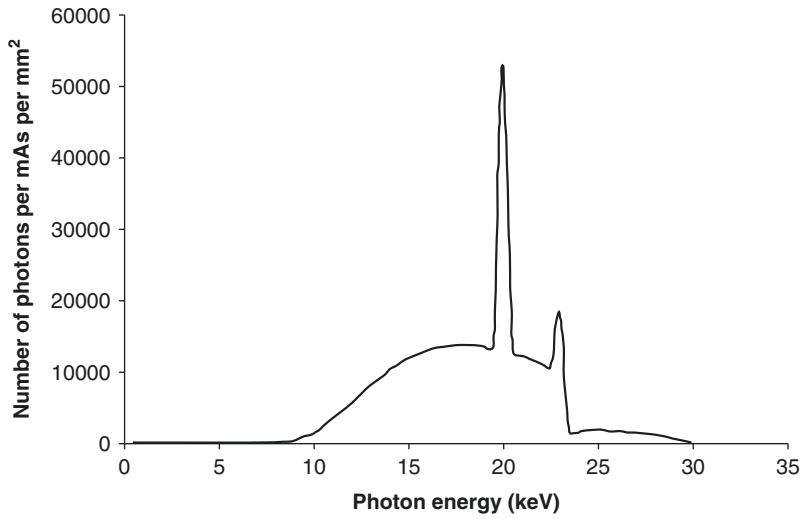

Fig. 18.2 X-ray spectrum for a typical rhodium target, rhodium filtered mammographic X-ray beam at 30 kV. The spectrum peaks around 20 keV due to the characteristic X-ray emissions of the rhodium target. This spectrum is suitable for imaging moderate-sized breasts

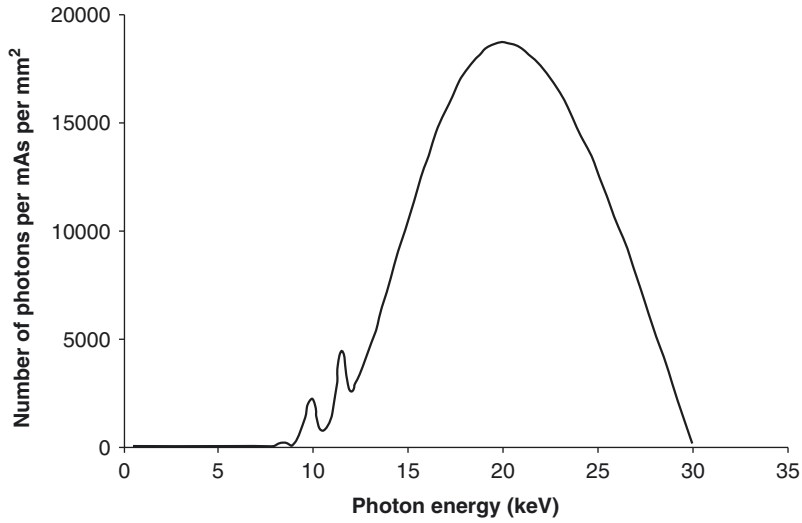

Fig. 18.3 X-ray spectrum for a typical tungsten target, aluminium-filtered mammographic X-ray beam at 30 kV

bination of target and filter material are used to maximize the number of X-ray photons in the beam that are close to the optimal energy. Since the introduction of digital mammography, sources with tungsten targets have become more common, and they have been paired with a variety of different filters. Fig. 18.3 shows the spectrum of a tungsten target, aluminium-filtered beam again at a tube voltage of 30 kV, with the X-ray tube again having a beryllium output window.

The shape of the spectrum is quite different from Fig. 18.2 even though the tube voltage is the same. The tungsten target has no K-characteristic X-ray peaks in this energy range, and the aluminium filter, which similarly does not have a K-absorption edge in this energy range, does not preferentially attenuate the higher energy end of the spectrum.

Compression Paddle Design

In mammography the breast is compressed using a rigid transparent plastic compression paddle that is motor driven. The use of compression

force reduces the thickness of the breast and holds it in place, which gives several advantages [7]:

- Reduced tissue superposition. Compressing the breast makes the tissues spread out over a larger area, reducing the possibilities for normal fibroglandular tissue masking pathology (reducing sensitivity) and of separate normal tissues projecting onto the image in a way that mimics a suspicious lesion (reducing specificity).
- Better spatial resolution. The breast is brought closer to the imaging receptor so that magnification and focal spot blurring is reduced.
- Reduced movement blur, even at the relatively long exposure times (1 s typical) common in mammography.
- Less scattered radiation in the image. The beam path length through the breast is shorter, so there is less material to do the scattering. Reducing the proportion of scattered radiation in the image improves image contrast.
- Lower radiation dose to the breast. Due to the same reduction of the beam path length through the breast, there is overall less absorption of X rays in the breast, so lower exposure levels are needed.
- Improved image uniformity. Compression spreads the breast tissue out more evenly across the image and makes pathology easier to detect.
- Diminished exposure time given the need for lower total exposure for an adequate acquisition [20, 21].
- The reduced path length makes practicable the use of lower energy (less penetrating) X-ray spectra. This gives greater subject contrast.

Compression in mammography is one of the few occasions in radiography where a technical advantage is gained without detriment to other aspects of the image, although there is a disadvantage in patient discomfort. Mammography systems measure the increasing amount of force resulting from a given small increase in compression to stop the motorised movement at a given compression. The compression force maximum limit set on mammography systems is 200 Newtons. A range of compression paddles are normally supplied with a digital mammography system to cover different types of projections [7]. Some typical types are:

- Flat rigid paddle - The basic flat paddle that covers the whole of the area of the digital image receptor and is used for full-field mediolateral oblique (MLO) and cranio-caudal (CC) views. The paddle maintains its shape parallel to the plane of the receptor and deforms only slightly when the compression force is applied.
- Tilting flat paddle - A flat paddle used for full-field MLO and CC views that allows rotation against a spring resistance so that during compression the chest-wall side of the breast will be thicker than the nipple side. The advantages are claimed to be that the design holds the breast in place more firmly.
- Sliding compression paddle - Suitable for imaging smaller breasts where the full area of the image receptor is not required. By sliding the paddle to one side or the other, the MLO view can be achieved using the edge of the breast support table to improve positioning.
- Spot compression paddle - This paddle has a raised cylindrical area that applies extra compression force over a small area. The advantages to spot compression are that better compression over the small area of interest is obtained, with all of the advantages above, but also that the spreading of surrounding parenchyma allows the outline of masses to be better visualised. Whereas features in superimposed tissue will spread out, mechanically harder malignant tissues will tend to retain their shape. Spot views are an additional examination often performed at assessment.

Magnification Compression Paddle

For magnification views, an add-on breast support table is used that raises the breast away from the plane of the image receptor by some 30 cm (depending on the magnification factor and

focus-to-receptor distance), so that the image is geometrically magnified. The compression paddle for this is smaller, as the X-ray field is smaller closer to the focus, and often has a step in the support arm to allow it to fix to the compression system at a point lower than the magnification support table. Please refer to Chap. 30.

Biopsy Compression Paddle

Various specialist compression paddles may be required for biopsy systems where the paddle has an aperture to accommodate the biopsy needle or device (Chap. 30).

Multiple advances have been proposed to lessen the uncomfortable aspect of breast compression, although the tolerance to it is variable (Chap. 16) while maintaining its advantages. Some such paddles have different levels of tilt, allowing for superior compression of the mid and anterior breast with less client discomfort. Another paddle has a curve in the center, more closely following the expected anatomy of the breast, especially in the CC view, allowing for superior compression towards the periphery of the breast compared to the central portion.

Another advance that does not involve only the re-design of the paddle is the use of a compression system that measures compression pressure (pressure = force/area) instead of force. Chapter 28 provides more details on this approach to breast compression. No recommendation is provided regarding the suitable compression force to consider the characteristics of the breast, namely compressibility, composition, and thickness. Several studies [22–27] investigated the best compression force in terms of dose, image quality, and client tolerance. One of these studies concluded that the amount of compression force has noticeable effects on image quality. Moreover, better image quality rates were consistently associated with higher compression forces [22], although the tolerance to compression is variable (Chap. 27). However, a study has also reported lower sensitivity at screening at the highest levels of compression force [28]. So, it is possible that the application of too high a compression level is not only detrimental to client comfort, but also to the clinical performance of the exam. Furthermore, another study evaluated the increase in compressed breast thickness, possibility for motion artifacts, and loss of tissue coverage if the compression force used was reduced by half, and no detrimental effects were found, while the level of comfort for the client was significantly increased [29]. Clearly, there is still room for further research in the area of breast compression.

Digital Mammographic Image Receptors

Digital image capture was first introduced into mammography as ‘small-field digital mammography’ for needle and core biopsy guidance, using detectors typically approximately 15 cm in size. Full-field digital mammography, with detector sizes up to the equivalent of the 24 × 30 cm was developed later.

The imaging advantages of digital mammography include a wide and linear dynamic range and the separation of the image capture and image display functions, so that the image display can be varied to optimally show the full range of recorded X-ray intensities [7]. This provides good visualisation of the skin line and nipple and has advantages when imaging dense breasts and younger clients [30]. Although a wide range of competing image capture technologies for digital mammography have been introduced since the introduction of digital detectors to this imaging modality, only two technologies, both consisting of full-field detectors, remain widely in use: the amorphous selenium-based direct and the scintillator-based indirect detectors [31].

General Features of Digital Mammographic Images

A digital image is not a continuous distribution of bright and dark but is composed of a finite number of points (or ‘pixels’), where each pixel has a value of brightness dictated by a stored numerical value. Digital images have the advantage that

they can be enhanced and manipulated by computer to extract the maximum amount of diagnostic information. Digital images can be stored, transferred, copied without detriment, and retrieved in a very efficient manner using computer mass data storage techniques [32]. They have the disadvantage of a limit to spatial resolution caused by the finite pixel size. Digital mammography receptors have a very linear response between pixel value and the radiation dose incident on the pixel over a very wide dynamic range, typically a factor of some 10,000:1. The choice of what dose is required for digital mammography is therefore driven by the signal-to-noise ratio required for a diagnostic image rather than a specific radiation dose to the receptor [7].

The Direct Digital Detector: Amorphous Selenium

In direct conversion detectors the X-ray interaction is converted directly to an electrical signal using an amorphous selenium (a-Se) layer, behind which lies an amorphous silicon microcircuit layer, which in turn is supported by a rigid substrate (Fig. 18.4) [7]. Selenium is a photoconductor, so is an electrical insulator in the dark, and a conductor when exposed to light or X-rays. The amorphous selenium is employed as a mammographic image receptor in the form of a thin layer (~0.5 mm) with a voltage applied between a large area electrode across the front surface, and an array of charge collection electrodes, one per pixel, on the back surface. These are linked to capacitors to accumulate the charge released during the exposure. These are linked, in turn, to thin-film-transistor switches to provide a line-by-line read out arrangement in which the charge stored for an individual pixel is passed pixel-by-pixel along the line until it can be measured by electronics external to the imaging sensor. Incoming X-ray photons interact photoelectrically in the a-Se layer producing electrons and 'holes' (the vacancy where an electron should be). Because of the high voltage gradient across the thin a-Se layer, the electrons move towards the positive surface electrode and the holes towards the negative charge collection electrodes. The electrons and holes do not move sideways as they must follow the direction of the electric field gradient, so image blurring from this source is minimal and the spatial resolution of the detector is good. At the end of the exposure, the charge signals (proportional to the radiation detected) from each pixel are read out via the thin-film-transistor switches and data lines. The charge signals are converted to digital values via charge amplifiers and with a digital-to-analogue converter and sent to the computer for assembly into an image.

The photoconductor is a layer of amorphous selenium that allows electrons to flow across it when exposed to X-ray photons [7]. The capacitor builds up a charge, proportional to the X-ray exposure for that pixel. The charge is transferred out of the device via the switch at the end of the exposure and converted to a numerical pixel value.

The a-Se layer has good photon capture characteristics in the mammographic energy range, and the lack of sideways spread of the electrons and holes carrying the image information allow the a-Se layer to be made relatively thick, resulting in an efficient detector. As the receptor is

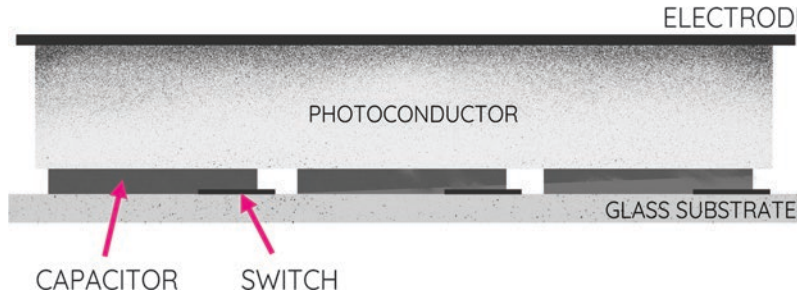

Fig. 18.4 Cross-section through a direct digital mammography image receptor

mounted rigidly in the breast support table of the mammography unit, it is always in the same position with respect to the X-ray beam, allowing the use of 'flat-fielding'. This is an important image calibration in which the receptor is exposed to the unattenuated X-ray beam under test conditions, so that variation in the X-ray intensity across the field and variations in pixel-to-pixel sensitivity can be removed from subsequent images. The removal of these fixed noise sources further improves the efficiency of the receptor.

The Indirect Digital Detector: Scintillator and Amorphous Silicon

Indirect digital mammography detectors use a two-step process for X-ray detection [6, 7, 18]. This type of receptor is similar to that commonly employed in digital radiography and consists of a thin crystalline scintillator layer closely coupled to an amorphous silicon microcircuit layer which is supported by a rigid substrate. Indirect conversion detectors work by first converting the incident X-ray distribution into a light image, then converting the light distribution into electrical signals addressable to a pixel location on the detector [6, 33, 34] (Fig. 18.5).

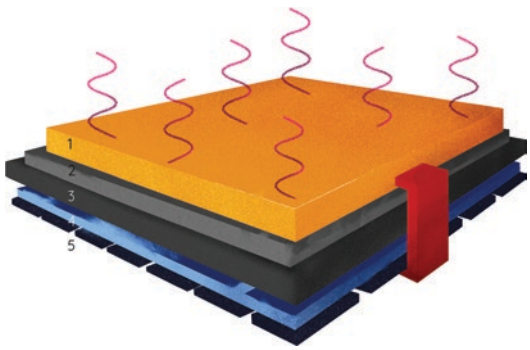

Fig. 18.5 Basic structure of detectors for digital mammography in integrated systems. *Layer 1*—detector material: CsI scintillator + transparent electrode or a-Se (amorphous selenium), *Layer 2*—a-Si array (amorphous silicon), *Layer 3*—base plate, *Layer 4*—driver board—readout board—driver board and *Layer 5*—glass substrate (Reproduced courtesy of Mário Oliveira)

The most successful scintillator is thallium-activated caesium iodide [7]. This has excellent X-ray absorption characteristics and can be grown in a channeled crystal structure that acts like a fibre optic guide to prevent light spreading sideways, giving to the detector improved spatial resolution. It is similar to the input phosphor material of X-ray image intensifiers. The scintillator layer is deposited onto an amorphous silicon micro-circuit array of light sensitive photodiodes and associated electronics to measure the signal from each photodiode. After the X-ray exposure is completed, a switching array of thin-film transistors and associated data lines allow the signals from the photodiodes to be fed out of the receptor array in sequence. These signals are then digitised and transferred to the computer to be assembled into an image.

This type of receptor is also mounted rigidly in the breast support table of the mammography unit, so the important flat-fielding correction described above can also be used, with the same removal of fixed pattern noise and resulting efficiency improvement [7].

The introduction of DBT and CESM resulted, for some systems, in some advances or modifications in these digital detectors. In the case of DBT, detector readout needed to be fast enough to allow for the acquisition of all the DBT projections within a short period. Although acquisition protocols vary substantially across DBT systems, some systems acquire 15 projections in ~2–4 s. Such acquisition protocol requires a detector readout rate that is much faster than what was needed for standard mammography. This rapid projection acquisition also required electronic advances to reduce detector lag, the latent signal from a previous acquisition that remains as a faint shadow in subsequent acquisitions [35]. For CESM, due to the use of a considerably higher-energy X-ray spectrum than that in standard mammography and DBT, the X-ray stopping layer, be it the a-Se layer or CsI scintillator, has to be thick enough to achieve a reasonable detection efficiency for these higher-energy photons [36].

Legacy Devices

Multiple detector technologies were used with different success in digital mammography [7]. Some of these, although not widely commercialised anymore, still have an important installed footprint, especially in countries where medical imaging equipment is not replaced at a relatively rapid pace. Probably the most common of these is computed radiography (CR) [7]. CR systems have been especially popular due to their being an affordable entry point to digital imaging. CR is used with traditional screen-film mammography systems, replacing only the screen-film cassette for the CR-based detector, which is then read out by a special scanner. As such, to introduce some level of digitisation to the mammography clinic, with CR it is not necessary to completely replace the imaging systems. As a result, CR systems continue to be prevalent in many countries with limited resources to allow for installation of full-digital systems. CR is based on the phenomenon of photo-stimulable luminescence.

When X-rays are incident on a material such as europium-doped barium fluorohalide, they produce high-energy photoelectrons that in turn produce ionisation that results in many lower energy electron–hole pairs. In conventional screen-film mammography, this happens in a screen in close contact with the film where the electron–hole pairs recombine to emit light that then exposes the film [7]. In photo-stimulable luminescence, however, less than 50% of the electron–hole pairs recombine, the others are trapped apart due to the presence of the doped sites in the phosphor. These electron traps are crystal lattice defects where halogen ion vacancies occur in the otherwise regular ionic lattice. These so-called ‘F’ or ‘Colour’ centres are created during manufacture by prolonged irradiation of the imaging plate with high intensity X-rays and ultra-violet light.

Following exposure, electrons can remain trapped at these defects for many hours or days, although the stored image gradually fades with time. The concentration of trapped electrons is proportional to the locally incident X-ray exposure. The electrons are trapped in this state until they are stimulated by light of a suitable wavelength in a CR plate reader, whereupon they are free to travel to the holes, recombine and emit light. The emitted light, which is linearly proportional to the locally incident X-ray intensity is then detected by a photomultiplier and digitised to form an image. The plate reader works by scanning an intense laser beam across the image plate on a line-by-line basis while the plate is slowly drawn through. A red laser is used to add enough energy to the trapped electrons to get them out of their traps and into the conduction band of the material. They can then move and recombine with a positive ion, dropping back to the ground energy state, and in doing so emit their excess energy as a photon of blue light. This weak light signal is picked up by a light guide and sent via a blue filter (to keep out the red light of the stimulating laser) to a photo-multiplier tube that measures the amount of light. This signal is then digitised to produce the raw ‘pixel value’ associated with that particular location on the image plate.

The scanning laser is focused to a diameter of approximately 0.1 mm to define the pixel of the image (although note that the imaging plate is continuous and not divided into physical pixels). Following read-out, the image plate is exposed to high-intensity light to completely erase any traces of the previous image, then reloaded into the cassette and ejected from the reader ready for reuse.

Because the CR cassette is not mounted rigidly in position, and several cassettes will normally be used in rotation, it is not possible to apply flat-fielding corrections in CR mammography, and the efficiency of the detector is reduced by the fixed pattern noise in the image arising from non-uniformity of the crystalline photo-stimulable phosphor. There is also an element of light spread in the phosphor from the read-out laser that leads to some blurring. Although, as mentioned, CR is a stepping stone to digital imaging, it has been found that CR is too inefficient, resulting in considerably higher doses and lower clinical performance than even screen-film mammography [37, 38].

A quite different type of digital mammography system that is no longer marketed is that employing a scanning fan beam of X-rays coupled to a moving one-dimensional photon-counting detector. This geometry is attractive in terms of its ability to reject scattered photons using a slit collimator at the detector, so no anti-scatter grid is required. The photon-counting detector, based on those used in high-energy experimental physics, counts each individual photon detected, and the pixel brightness is dictated by the total photons counted during the time the X-ray beam was swept over the pixel position. This has the advantage that low-level fluctuations caused by thermal excitation in the amplifiers and electronics can be rejected, leaving only the higher energy photon counts, so one source of image noise can be negated. In addition, later generations of these detectors included more advanced electronics allowing for the energy of the detected photon to be determined, and therefore photons of different energy levels could be discriminated.

Although this energy resolution is limited, it does allow for the simultaneous acquisition of at least two images, one consisting of the detected low-energy X rays and the other only including the high-energy X-rays. This is ideal for CESM acquisitions. However, the motorised movements of the scanning beam are complex, the X-ray tube loading tends to be high, and the scan time is generally longer than the exposure time for a two-dimensional receptor. These issues, among others, resulted in this technology, for now, not being pursued further for mammography.

The Automatic Exposure Control System

In the 1980s the automatic exposure control (AEC) system was implemented in mammography equipment with the aim to provide the most optimal optical density on the film. For the most part, current mammography AEC systems adjust the exposure time, while the tube current is often fixed or varied only across very few settings. Mammographic AECs can be very sophisticated, making allowance for the attenuation of the breast, the energy of the beam, and able to automatically select not only the exposure time, but in some implementations also the target/filter combination and tube voltage for the breast being imaged. For this, most current digital mammography systems use part of the digital detector itself as the AEC sensor. The image acquisition starts with a short low-dose exposure, called pre-exposure or scout image, of the breast and the resulting signal is sampled automatically to identify the area with the lowest signal level, which usually corresponds to the densest areas of the breast. This information, in combination with the compressed breast thickness obtained from an electronic readout of the position of the arm of the compression paddle, is used to select the optimal image acquisition settings.

In some systems, the breast thickness alone determines the target/filter (for systems that have more than one of either) and the tube voltage to be used, while the pre-exposure is only used to set the tube current-exposure time product. In other systems, all these image acquisition settings are determined based on the pre-exposure [7]. The most sophisticated AEC systems can detect non-breast tissue on the pre-exposure image, so the AEC can be used when imaging breasts with implants or some medical device.

The AEC system is usually programmed to provide a constant contrast-to-noise ratio with increasing compressed breast thickness, although at the largest thicknesses the signal-to-noise ratio is allowed to decrease slowly, to avoid the largest increase in doses for the largest breasts.

Optimisation of Digital Mammography

The linear response and wide dynamic range of digital mammographic receptors means that images can be successfully acquired over a large range of doses [7]. This provides several possibilities for image optimisation and dose reduction, but equally also allows systems with sub-optimal setups to acquire images at higher patient doses than are necessary. The phenomenon of ‘exposure creep’ has been identified in general digital radiography, where average

patient doses can rise due to the natural human inclination to make the images look better, and the fact that images are not rejected for being 'too good'. Modern automatic exposure control software may offer alternative combinations of automatic exposure factors that either optimise for contrast (at the expense of higher dose) or dose (at the expense of poorer contrast-to-noise ratio).

Given the sophisticated AEC systems now available, after adequate installation and commissioning, there are few instances, if any, for which it is justified to perform mammographic acquisitions with manual settings. As mentioned, some systems do not have AECs sophisticated enough to perform correctly in the presence of breast implants or implanted medical devices, and therefore the use of manual settings in this case is justified and necessary.

Display Devices

With a pixel size of 0.05–0.1 mm, and a typical field size for full-field digital mammography of 24 × 30 cm, a digital mammography image may well be composed of over ten million pixels [7]. Specialist medical-grade display monitors are required to provide an adequate display for primary reporting. Lower specification displays may be used as 'review' monitors in the mammography room for the practitioner to confirm the quality of image acquisition, but these should not be used for primary reporting. By now, it is exceedingly rare to see cathode-ray-tube-based digital mammography monitors, all of them having been replaced by LCD flat-panel displays.

Although there can be some flexibility in the format of simultaneous image display for reporting, in general two high-resolution (approximately 2000 × 2500 pixels = 5 megapixels) monitors in portrait orientation will be required for a reporting workstation as usually two images need to be compared. Over the last few years, 10 to 12 MP monitors have been introduced, so one monitor, in landscape mode, is used instead of two 5 MP portrait monitors. An additional low-resolution monitor may be required to display patient information, worklists, and other textual diagnostic reports.

It is not generally expected that the display monitor will be capable of displaying the full resolution of the recorded image as a complete frame, but that magnification, pan, and zoom within the image will be used to display all pixels when this is needed. With current 5-megapixel monitors, only a proportion of the breast image can be displayed at full resolution.

An important distinguishing feature of medical-grade displays is their maximum luminance. Ideally this should be 450 cd/m^2 or higher (much brighter than standard computer displays) so that a large ratio between maximum and minimum can be maintained, and susceptibility to the effects of ambient lighting is reduced. Careful consideration to the design of the viewing room is still required, however, as the brightness of the monitor itself will light up the room (as well as more obvious light sources such as open doors and windows) and structured reflections of room surroundings and indeed the observer superimposed on the viewed image will reduce its contrast and may introduce distracting features. Since the introduction of DBT, tomosynthesis-ready mammographic monitors have been developed, which support a rapid enough refresh rate to allow for the quick scanning through the different reconstruction planes without any motion blur.

The DICOM Greyscale Standard Display Function

DICOM is the medical image interchange standard that allows varied imaging-related systems to communicate images and image-related information, such as radiation dose structured reports. One element to DICOM that is particularly important from the radiological reporting standpoint is the Greyscale Standard Display Function (GSDF). This is based on a psychophysical model of the human visual system and is designed to maximise the number of 'just noticeable differences' that a given display can reproduce, and to give a perceptually linear greyscale, with the same small change in contrast visible in a dark part of the image as in a light part. Usually, the GSDF boosts the signal in the lighter side of the greyscale

range. If the GSDF is correctly implemented for a given monitor, it should give the best display that monitor is capable of in the viewing conditions where it is used. The GSDF attempts to make the best of the display's capabilities but cannot make a sub-par display in poor viewing conditions as good as an expensive megapixel grey-scale monitor in good viewing conditions.

Display Tools

Display workstations would be expected to provide a user interface providing an efficient throughput of images and a range of display tools, typically including:

- Magnification, zoom, and pan (roam)
- Contrast and brightness adjustment (windowing)
- Image flip and rotation
- Black/white inversion
- Spatial measurement
- Edge enhancement and noise reduction (spatial frequency filtering)
- For DBT: manual scanning as well as automatic cine mode through the DBT slices

Some of these features are further explained below.

Windowing

Post-processing of digital images by windowing is a very powerful feature of digital imaging that also applies to CT, MR, and radioisotope imaging [32]. As the brightness of a pixel is dictated by an integer number (the 'pixel number') in a digital image, there are a finite number of values that the brightness level can take. Digital mammography systems might typically digitise to 12 bits (4096 grey levels), whereas the display monitor will probably only have a capability of displaying 1024 levels of luminance (10 bits). In addition, the human visual system is only capable of distinguishing about 100 grey levels at one time in an image, even under ideal viewing conditions, so it follows that if all the information present in a digital image was displayed on the monitor at once, small differences in contrast, although recorded successfully, would not be distinguishable.

The solution to this problem is to display only a selected range of pixel values, thus increasing the displayed contrast for that subset of levels. This 'window' of pixel number values is defined by a window 'width' and window 'level' (Fig. 18.6). By altering the display window width and level settings, the observer can optimise the display of the range of grey levels for the diagnostic task being undertaken, and any contrast

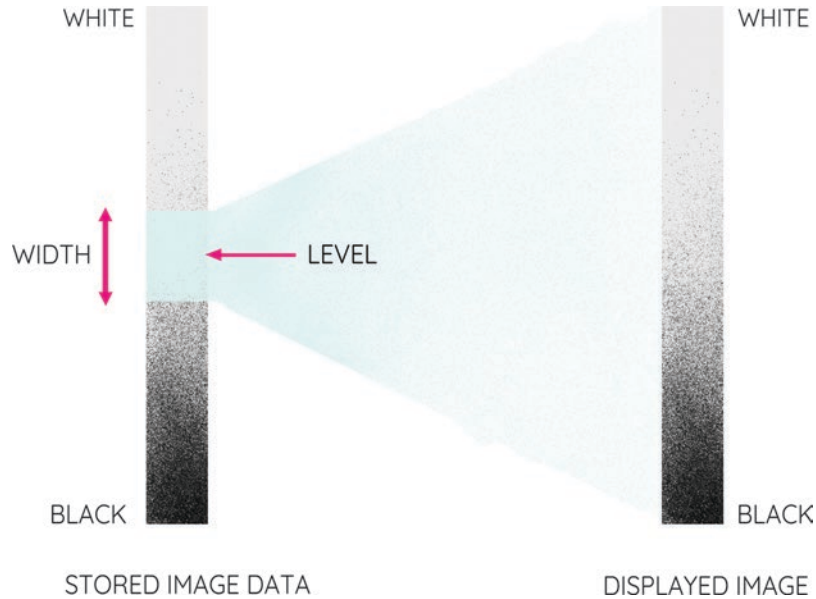

Fig. 18.6 Diagrammatic representation of image display windowing. The display width and level define a subset of the stored image grey levels which is expanded to fit the full luminance range of the display device

recorded in the image can be displayed, but the time taken to make many such adjustments can become a factor in reporting high volumes of images. Therefore, preset window levels are usually set and used when a mammographic image is displayed, while the user interface for window width and level adjustment is usually quite intuitive, using a dedicated keypad, in addition to the computer mouse or trackball.

Spatial Frequency Filtering

Images can be thought of and analysed as sets of spatial frequencies. In general, low spatial frequencies are associated with uniform greyness or slowly changing gradients, whilst high spatial frequencies are associated with sudden changes in brightness such as at sharp edges or patterns of dots or lines [32]. By applying a spatial frequency filter, ranges of spatial frequencies can be enhanced or attenuated. Enhancing high spatial frequencies enhances the contrast of sharp edges, e.g., calcifications and linear structures, and generally 'sharpens' the image. Unfortunately, high frequency enhancement comes at the price of also boosting the noise that lies in this frequency band, so subtle enhancement is the most effective.

Attenuating high frequencies effectively blurs the image, and this can be used to reduce the appearance of quantum noise in some situations. Various layers of image processing, including spatial frequency filtering, are routinely used in digital mammography. Whilst this processing can make improvements to clinical images, it can also cause problems with quality control phantom images, for which the image processing often must be deselected.

Quality Control of Display Devices

Monitor performance reduces with age, and regular quality control checks are required. Regular user checks should include the systematic visual checking of a test pattern, such as the SMPTE pattern, the AAPM TG18 pattern (Fig. 18.7), and/or the more recent AAPM TG270 patterns.

Images of a suitable pattern should be accessible from the reporting workstation and for review monitors. Quantitative tests of the monitor performance, which include measurements of luminance over a range of grey levels, and assessment of the number of 'just noticeable differences' that the monitor can deliver in the lighting conditions where it is used. It is common for medical-grade monitors to support self-calibration, where the monitor makes measurements of its own luminance output and adjusts its calibration accordingly. The calibration takes account of room lighting conditions, so problems can arise if the lighting in the room at the time of self-calibration is not the same as when it is used for reporting.

A systematic review of viewing conditions and monitor specifications in mammography was performed in 2020 by Papathanasiou and colleagues [39]. This review provides an overview and further information which would be useful for the reader.

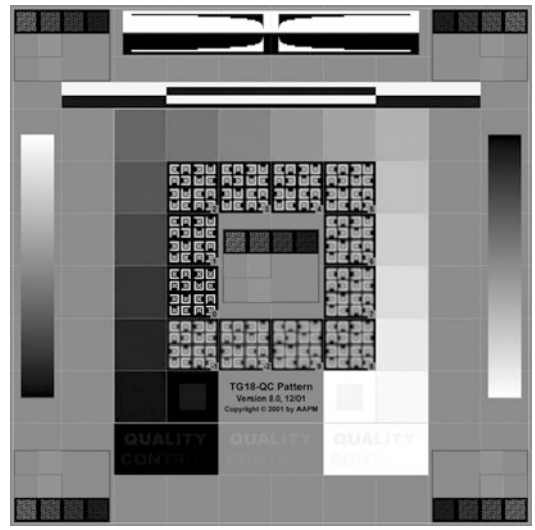

Fig. 18.7 A common quality control test object for display monitor testing. This is the AAPM Topic Group 18 (TG18) test pattern. The pattern features grey scale, image alignment, high spatial resolution and low contrast tests

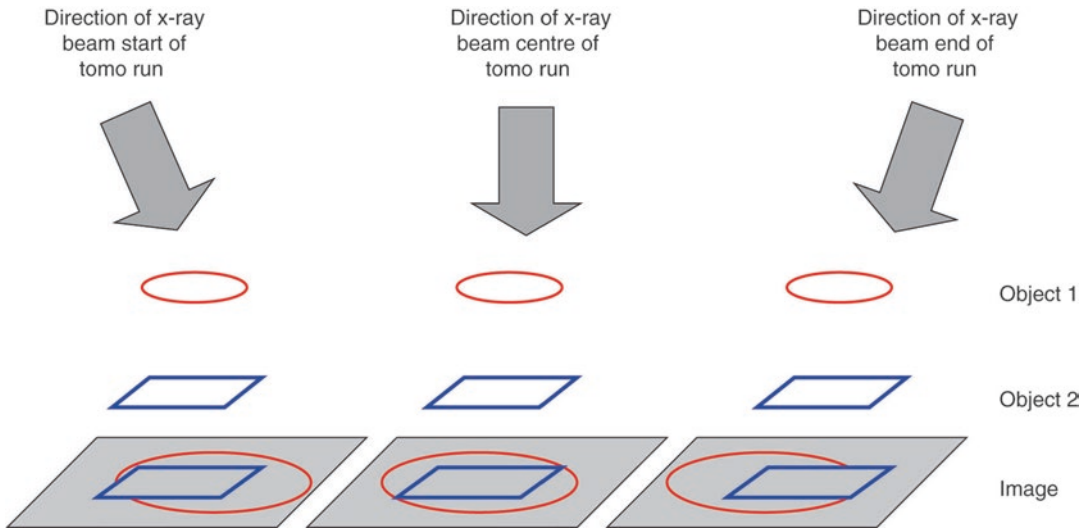

Fig. 18.8 Tomosynthesis image acquisition - The breast is held compressed against the stationary support table, and a sequence of small exposures is made as the tube gantry moves through an angle

Digital Breast Tomosynthesis

The use of MLO and CC views in standard mammography ameliorates, to some degree, the impact of tissue superposition on sensitivity and specificity. However, acquiring two planar images of the breast cannot replace obtaining the more ideal image, which would be a 3-D array of X-ray attenuation, from which it would be possible to display any desired image plane. DBT falls some way short of that ideal but does provide some useful depth information. This benefit, combined with the ease of introduction of DBT due to it sharing the same imaging platform as mammography, has resulted in it being quickly taking up for X-ray breast imaging in the clinic. Further information on tomosynthesis is available in Chap. 37, some aspects are summarised below.

Image Acquisition

For DBT acquisition, the breast is compressed against the image receptor as normal, but instead of one exposure with the X-ray central ray orthogonal to the image plane, a sequence of shorter exposures is made as the tube gantry moves through an arc. The result is a series of images, taken with the source of X-rays stepping through the swing angle that can range from ±7.5° to ±25° of the normal vertical position. The projections will be subtly different, as the X-ray shadow of objects close to the top of the breast will appear to move relative to the image frame as the X-ray focus moves, but objects close to the support table will be imaged in the same place (Fig. 18.8).

Reconstruction

To produce the tomographic image, the series of projections must be reconstructed into a single image that emphasises features at a particular depth within the breast. In the simplest form of tomosynthesis, this could be done by shifting the projection images with respect to the image frame, so that the features at a selected depth all appear in the same place within the frame. These shifted images are then added together. The addition reinforces the contrast of features in the selected plane, where they are in the same position, but tends to blur out objects in other planes.

The degree of blurring (or technically streaking, as the blur occurs in the direction of X-ray

tube movement) increases with distance from the selected plane. The result of image reconstruction is therefore an image reminiscent of film-screen tomography, where the observer can focus on objects in the intended image plane but tends to 'see through' the blurred features in other planes. This is distinct from true tomography (e.g., CT), where each image is a true cross-sectional cut through the object with no overlying or underlying structure. More sophisticated DBT image reconstruction methods based on filtered back projection (a variant of CT reconstruction) or iterative techniques are used in commercial DBT designs, but because of the very limited range of angles at which the projections in DBT are recorded, these still cannot recover enough information to produce pure tomographic slices.

Image Interpretation

To create an image stack suitable for mammographic reporting, the tomosynthesis reconstruction process is repeated with the calculated in-focus plane shifted, typically, one millimetre down from the previous one [9]. This cycle is repeated to eventually produce an image stack of perhaps 50 or 60 tomosynthesis images, for a typical compressed breast thickness. The entire stack of tomosynthesis images can be reconstructed from just the one set of projections. It should be noted that the information contained within one of these in-focus planes pertains to a lot more than 1 mm of tissue, but rather ~5 to 10 mm of breast tissue content, or more, depending on the contrast of the features, can appear in the plane being depicted.

To report the images, the viewer controls the selected in-focus plane shown on the display screen, and this can be rapidly swept up and down through the image stack [40]. The act of stepping through the images on the display allows the observer to build up a 3-D impression of the relative positions of features within the volume of the breast. For example, a small detail feature, such as a cluster of calcifications, will gradually come into sharp focus as the displayed in-focus plane approaches its true depth, then will fade out of focus as displayed image moves beyond it.

Radiation Dose for Tomosynthesis

The radiation dose to the patient from DBT would be expected to be marginally higher than for conventional 2-D views, because the X-rays forming the projection views at the extremes of the angular swing have to traverse a greater thickness within the compressed breast. Most commercial implementations aim to keep the dose for DBT comparable with conventional 2-D views.

Synthetic Mammograms

Acquiring a standard mammogram in addition to a DBT would result in, at least, doubling the dose of an exam. To avoid this, the information acquired in the DBT projections can be used to reconstruct not only the DBT pseudo-3D image, but also a single planar image, the synthetic mammogram. This planar image is useful to both compare the current exam to priors, and to provide the interpreting radiologist with an overview of the breast, so that they can determine if there is an area they want to focus on, in addition to the review of the entire image stack.

There are two approaches to creating this synthetic mammogram. In the first place, the aim could be to create an image that is as close to what the real mammogram of that imaged breast would be [41]. Another approach is to create a planar image that includes all the interesting features found in the DBT image, and nothing more [42]. Of course, this latter approach is much more challenging, and requires some level of automated interpretation of the DBT image to identify the features of interest. With the advent of artificial intelligence-based image interpretation methods, we can expect that this latter approach to synthetic images will become more common.

Dedicated Breast CT

Dedicated breast CT (BCT) is a newer technology that delivers fully tomographic images of the breast [11–13, 43]. To achieve this, BCT uses the same principles as conventional (body) CT, but it is optimised, in terms of acquisition geometry, X-ray source, and detector, to imaging the breast. In addition to its true tomographic capabilities, BCT has the advantage that it does not involve any breast compression, increasing client comfort.

From the equipment point of view, BCT involves a horizontal gantry involving an X-ray source and a digital X-ray detector, both rotating around a vertical axis. This gantry is placed under the client table, on which the client lies prone, with the breast to be imaged pending through a hole on the table. The vertical axis of rotation of the gantry is located at the center of the hole for the breast.

To date, there are two implementations of BCT: cone-beam BCT and spiral BCT [11–13, 43]. In the former, the whole breast is included in each projection, and the detector is an energy-integrating flat-panel detector, similar to the indirect detectors described above for mammography. A single cone-beam BCT image consists of the acquisition of 300–500 projections during an entire 360° revolution around the breast, in 10–16 s. In the current implementation of spiral BCT, a fan-beam of x-rays and a narrow photon-counting detector strip are used, and these two revolve around the breast while also translating downwards. In this implementation, the detector behaves similarly to the photon-counting detectors that are described above as previously having been used in some mammography systems.

BCT has many advantages over mammography and DBT. It does not involve breast compression and results in a fully tomographic image. In some implementations, the dose for a BCT acquisition is equivalent to that of the acquisition of two views of mammography, resulting in the same dose for a complete exam of one breast. In addition, presenting a calcification cluster in its true three-dimensional distribution should make characterising it easier for an interpreting image reader. In addition, for low-contrast soft-tissue masses, the tomographic nature of BCT should make them easier to detect and characterise. Finally, since BCT is performed with higher X-ray energies than mammography and DBT, it is ideal for contrast-enhanced imaging, with the spectra being very well tuned for iodine imaging [13].

However, the installed clinical, and research, base of BCT is, for now, much smaller than that for DBT and CESM, so the clinical evidence of its performance and its most suitable clinical use is still lacking. Since, as opposed to DBT and CESM, BCT requires a completely new system, its acceptance has not yet been as widespread as that of these other two modalities. Therefore, it remains to be seen what, if any, will be the clinical impact of BCT.

Contrast-Enhanced Spectral Mammography

As mentioned, CESM has been introduced over the last few years to breast clinics with the aim of providing functional information in addition to the anatomical information obtained with standard mammography. Just like DBT, CESM shares the same platform with standard mammography with only relatively minor modifications, and now the typical mammography system can perform the three types of imaging [17]. In fact, the modifications to a mammography system needed for CESM are fewer than those needed for DBT [44]. Usually, mammography systems already were able to use tube voltages of up to 45 or 49 kV. Therefore, only specific additional filters, such as copper or titanium, were needed to allow for adequate high-energy image acquisition. As discussed previously, the use of higher-energy x-rays could also necessitate a thickening of the X-ray absorption layer of the digital detector, to ensure an adequate stopping power, avoiding a substantial increase in dose.

To obtain the recombined, iodine-only, image, in CESM two images are acquired of the breast during one compression event: a low and a high energy image. The low energy image is the equivalent of a standard mammogram. The high energy image is acquired immediately thereafter, with

the higher tube voltage, ~45–49 kV, and an additional filter. The aim of this higher energy spectrum is for it to include all or most of the photons above the k-absorption edge of iodine, which is 33.2 keV.

At this energy, the rate of absorption of x-rays of iodine increases substantially, while breast tissue does not have any k-edge present within the mammographic spectrum range. Therefore, in the presence of iodine, a specific mathematical combination of the low and high energy images results in an image that depicts only the iodine. Since the iodinated contrast agent is mixed in the blood, the recombined CESM image depicts the distribution of blood in the breast and in any pathology present. Due to the chaotic nature of blood vessels in the tumor growing rapidly through angiogenesis, blood leaks out from the tumor vasculature, resulting in tumors appearing enhancing in these images. More information on CESM can be found in Chap. 38.

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Appendix

Test your learning and check your understanding of this book's contents: use the "Springer Nature Flashcards" app to access questions using https://sn.pub/dcAnWL.

To use the app, please follow the instructions in Chap. 1.

Flashcard code: 48341-69945-ABCB1-2A8C7-CE9D2.

Short URL: https://sn.pub/dcAnWL.

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