Clinical Mammographic and Tomosynthesis Units

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23.1 Introduction

Mammography is an X-ray examination of the breast, obtained by dedicated equipment, optimized for breast cancer detection or characterization. Although introduced as early as the 1970s, mammography is still considered the most effective imaging technique for early detection of breast cancer, and it is now widely used as a screening and diagnostic tool. A major challenge is to image subtle features, like small, low-contrast masses or microcalcification clusters, within an X-ray image of the breasts (inherently low-contrast), allowing breast radiologists to make a correct interpretation. Since the 2000s, mammography has gained an advantage from the evolution of image detector technology, moving from screen-film combinations to digital detectors. The major benefit of digital mammography comes from the separation between image acquisition and image display; post-processing tools became accessible to the users on dedicated workstations, making radiologists more confident about their image interpretation. Non-inferiority of digital versus screen-film mammography was firstly proved by the Oslo 2 Trial (Skaane and Skjennald 2004), while the publication of the results from the Digital Mammography Imaging Screening Trial (DMIST) showed that digital mammography is more accurate than screen-film mammography for young women and women with dense breasts, that is, for those groups of women for whom screen-film mammography was less sensitive (Pisano et al. 2005).

More recently, digital mammography advances (see Section II, Chapter 19 of this book) allowed us to face the main limitation of projection imaging, namely the tissue superimposition and the associated reduction of cancer detection, as well as a certain number of false positives, by developing other applications like digital breast tomosynthesis (DBT) (see Section II, Chapter 20 of this book) and contrast-enhanced spectral mammography (CESM, also called CEDM, contrast-enhanced dual-energy mammography).

In the following, the main components of clinical mammography and tomosynthesis units will be described, and both the aspects of diagnostic performance and radiation dose will be discussed.

23.2 Components of Clinical Mammography Units

A schematic of a clinical mammography unit is illustrated in Figure 23.1, where the multiple components can be grouped as follows:

- X-ray source
- Acquisition geometry
- Digital detector
- Scatter rejection methods
- Automatic exposure control
- Image post-processing
Mammography X-ray tubes always have rotating anodes, and are equipped with dual filaments to produce two focal spots of different nominal size, 0.3 mm to acquire standard mammography images (contact views), and 0.1 mm to acquire magnification views. The tube port is usually a thin beryllium layer (0.5–1 mm), substantially transparent to X-rays (absorbs only photons with energy lower than 5 keV).

Tube voltage, obtained by high frequency generators (Sobol 2002), is another parameter which differs significantly between mammography and conventional radiography: the typical range is between 25 and 35 kV for mammography, against the 50 to 150 kV range of radiography. This is due to the need of maximizing the limited contrast produced by the different breast tissues, whose small differences in absorption coefficients slightly increase at lower photon energies (i.e., lower tube voltage values).

23.2.2 Acquisition Geometry

Mammography X-ray tubes are usually positioned to maintain a source-to-image receptor distance of about 65 cm. As the X-ray beam must be sufficient to include the area covered by the breast, without exposing other parts of the patient body, the tubes are physically tilted (as shown in Figure 23.1) to cover about a 24 × 30 cm² field of view (FOV).

The intensity of the X-rays emitted is not uniform in the cathode-anode direction, but the number of emitted photons decreases on the anode side compared to the cathode side, because of the heel effect (Andolina and Lille 2010). Thereby, the X-ray tube is positioned with the cathode corresponding to the chest wall of the patient and the anode to the nipple area, to match the shape of the breast; in fact, despite the breast compression, the posterior part of the breast, close to the chest wall, remains thicker, requiring more photons, while the anterior part, close to the nipple, is thinner, and, consequently, a lower X-ray intensity is sufficient to produce a good image.

As previously mentioned, the breast is compressed during mammography examination. Breast compression is fundamental to obtain a good mammogram, reducing breast thickness and tissue superimposition, and limiting accidental breast motion during the exposure, which would cause image unsharpness. It also reduces the amount of radiation dose necessary to obtain the image, and the scattered radiation fraction. In the early days of digital mammography, it was announced that breast compression (which is not judged positively by women undergoing mammography) could be reduced compared to screen-film mammography, because of the post-processing; however, this was never proven, and good breast positioning and firm compression are still key points to obtain high quality, diagnostic images. Breast compression is performed by a compression paddle, usually a thin polymethyl-methacrylate (PMMA) or polycarbonate plate, which moves (usually) parallel to the breast support using a motor-driven system. The radiographer uses his/her hands for breast positioning and his/her foot to apply the appropriate compression; a final manual adjustment is also possible by mechanical knobs placed on both sides of the paddle assembly. The size of the compression paddle fits the detector field-of-view, typically 18 × 24 cm² or 24 × 30 cm² for “contact views” (obtained with the breast compressed on the breast support, i.e., as close as possible to the image detector). Additional compression paddles can

FIGURE 23.1 Schematics of a clinical mammography unit, illustrating all the components described in the following sub-sections.

23.2.1 X-ray Source

The X-ray source includes all the X-ray tube parts and the additional filters.

When screen-film technology was used as image support, the anode target material was one of the major differences between the X-ray tubes used in mammography and those used in conventional radiography. Molybdenum (Mo) was the anode target material mostly used in screen-film mammography, producing spectra with a significant amount of characteristic radiation photons (energy peaks at 17.5 and 19.6 keV) and a continuous bremsstrahlung spectrum, whose maximum energy value in keV is equivalent to the tube voltage in kV (Beaman and Lillicrap 1982). Similar spectra, slightly more penetrating, were produced by rhodium (Rh) anodes (energy peaks at 20.2 and 22.7 keV). Additional filters are always employed in mammography, with the aim of shaping the X-ray spectra by removing the lowest and highest photon energies (which do not constructively contribute to image formation). Filters typically used with Mo targets were a 0.030 mm Mo filter or a 0.025 mm Rh filter; the 0.025 mm Rh filter was used in combination with Rh targets. Spectra produced by Mo/Mo, Mo/Rh, or Rh/Rh anode/filter combinations suitably matched the exposure latitude of screen-film detectors. However, the introduction of digital detectors, with wider exposure latitude and post-processing, which reduced the need of maximizing subject contrast during image acquisition, was accompanied by the increased use of tungsten (W) target material, the same material used in conventional radiography, characterized by a higher melting point, and consequently higher maximal tube loading. Nowadays, most clinical mammography units use a single W anode target, filtered with Rh (0.05 mm), Ag (0.05 mm), or Al (0.7 mm), depending on the system.

In Figure 23.2, spectra produced by Mo/Mo, Rh/Rh, W/Rh, and W/Al anode/filter combinations, all at 30 kV, are compared.

A few digital mammography systems still mount tubes with two anode tracks: Mo and Rh targets for General Electric (GE) systems, and Mo and W targets for Siemens systems, even if they use almost exclusively Rh and W, respectively.
be provided for “spot views” (the breast is still compressed on the breast support, but compression is applied to a limited portion of the breast to produce a good image of the area under investigation). Magnification views can also be acquired by mounting an additional support which permits one to compress the breast halfway between the X-ray source and the image detector; the area of the breast under investigation is then projected onto the image plane with a magnification factor of about 2. Magnification views are usually used to enlarge possible microcalcification clusters. Figure 23.3 shows the three different types of breast compression/image acquisitions already described: contact, spot, and magnification views.

In digital mammography units, the X-ray beam collimation is automatically set by the compression paddle, and the type of mammographic view selected on the acquisition workstation console.

A standard mammography exam includes two contact views of each breast, with the breast compressed in cranio-caudal (CC) and medio-lateral oblique (MLO) position, respectively. In CC view, the gantry is vertical, with the X-ray tube perpendicular to the floor. The breast is compressed on the breast support, with the chest wall leaned against the detector side, and pulled anteriorly to have the nipple in the image profile. The pectoral muscle is not always visible in CC views. In MLO view, most of the breast tissue is visible, including the pectoral muscle. The gantry and the detector are angled (approximately ±45°) and the breast is compressed from the upper medial direction (Linfors and Petress 2012). Extra-views with different geometry may be acquired in the case of suspicious findings or diagnostic doubts, in a diagnostic workflow.

An example of mammography examination is depicted in Figure 23.4.
23.2.3 Digital Detectors

From the beginning of digital mammography until now, several changes and advances have been applied to clinical systems. However, the general scheme illustrated in Figure 23.5 can be analyzed to describe all the digital technologies that have been used and are currently used in commercial equipment.

The first distinction can be made between integration and photon-counting detectors, although only one system is available on the market using a photon-counting detector compared to several systems with integration detectors. Integration detectors are characterized by an active area covering the full field-of-view; for this reason, systems employing this type of detectors are often called “full-field digital mammography” (FFDM) systems. Furthermore, integration detectors take their name from the way they produce the signal-to-noise ratio (SNR) in each detector element, resulting in the final image. In fact, a SNR proportional to the intensity and energy of the photons transmitted through the breast is accumulated during the entire exposure time, and the resulting image is a “transmission map” produced by the accumulated photon energy delivered at each detector element over the exposure time. Different sources of noise contribute to the final SNR of integration detectors.

In contrast, photon-counting detectors can detect the individual photons interacting with the detector material, producing a signal above a preset threshold in the detector elements (see Section I, Chapter 13 of this book). This permits one to eliminate the contribution of noise sources other than the quantum noise, by setting the discrimination threshold above the noise level. Photon-counting detectors need fast electronic read-out, with an independent channel for each detector element. For mainly economic reasons, photon-counting detectors are not yet full-field, but have a narrow slot area, and the full-field coverage is guaranteed by scanning. The unique clinical mammography system with photon-counting detector (silicon microstrips) was originally developed by Sectra, and successively purchased by Philips. Its strength is the capability of keeping the radiation dose level very low, approximately 40% below the dose level of the FFDM systems (Cole et al. 2012).

Most FFDM systems employ flat panel detectors (FPDs). There are two types of flat panels for X-ray imaging: (1) FPDs with a scintillator layer to convert the incident X-rays in light photons,
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23.2.4 Scatter Rejection Methods

When an X-ray beam interacts with body tissues, it is partially absorbed and partially transmitted. The transmitted X-rays unavoidably include both primary and scattered radiation. Primary radiation, associated to the photoelectric absorption, produces a useful signal for the image formation. In contrast, scattered radiation, mainly caused by Compton interactions, can be considered as an additional random noise that degrades the final image quality. The impact of scattered radiation on image quality is traditionally described by a parameter called “contrast degradation factor” (CDF), because, when images were acquired on films, the major effect of scattering was the reduction of image contrast (Rezentes et al. 1999). CDF was defined as the ratio between the transmitted contrast in the presence of scatter, \( C_s \), and the transmitted contrast in absence of scatter, \( C_0 \), according to the formula

\[
CDF = \frac{C_s}{C_0} = \frac{1}{(1 + SPR)}.
\]

where SPR was the scatter-to-primary ratio (Neitzel 1992). In digital mammography, scatter increases image noise and reduces contrast, affecting the signal difference to noise ratio (SDNR).

In general, the amount of scattered radiation is strongly dependent on breast thickness, and weakly dependent on X-ray beam energy (at least for energies used in mammography). For example, a breast of 5 cm compressed thickness produces a SPR approximately equal to 0.5 (Boone et al. 2000), which corresponds to a CDF = 1/1.5 = 66.7%. This shows that it is necessary to introduce some type of device or a method to reduce the degradation of image quality caused by the scattered radiation.

Figure 23.6 shows the two standard methods used to reduce the scatter effect and improve mammography image quality. On the left, the method normally used in screen-film, CR, and FFDM mammography, introducing an anti-scatter grid inside the X-ray source.

**FIGURE 23.6** The two methods to reduce scatter effects in mammography. (Left) Moving focused grid, used by any type of FFDM unit. (Right) Scanning slot system, used by the photon counting equipment.
breast support, between the breast output and the image detector; it is a moving, focused grid (a series of lead septa, spaced by radiotransparent material), which intercepts preferably scattered photons. On the right, the method used by the photon counting digital system: scatter rejection is performed by using a narrow X-ray beam at the breast entrance, and further collimating the output beam to fit the slot detector size, while the narrow detector performs a breast scan in the left-to-right direction (Aslund et al., 2006). Recently, Siemens have implemented computerized scatter removal.

### 23.2.5 Automatic Exposure Control

The automatic exposure control (AEC) is the device used to control the exposure factors, acting on both image quality and radiation dose. Anode/filter (A/F) combination, kV, and current-exposure time product (mAs) must be selected on the basis of the effective breast absorption, to provide diagnostic image quality, while keeping the dose as low as reasonably achievable, according to the radiation protection principles. In screen-film and CR mammography, the AEC was constituted by one or more dose sensors (usually ionization chambers or semiconductor diodes) located under the breast support (Figure 23.1). The sensor(s) integrated the X-ray signal transmitted through the breast during the exposure time, until the accumulated signal reached a preset threshold value; the achievement of the threshold terminated the exposure. In DR mammography, the need for additional sensors as AEC is no more required, and the digital detector used to acquire the image, or a portion of it, is normally also used for automatic exposure control.

In general, the AEC adjusts the detector air kerma (using the technique factors A/F, kV, and mAs) for the given breast absorption, with the main purpose of maintaining high image quality. The mean X-ray energy is increased (by increasing tube voltage and/or using more selective filtration) as the breast thickness increases, while the tube load (mAs) is adjusted to provide a constant signal or signal-to-noise ratio to the image detector. AECs can usually operate in three modes: (a) full automatic mode, (b) semi-automatic mode, and (c) manual mode. In full automatic mode, the AEC automatically selects all the technique factors (A/F, kV, mAs), on the basis of breast absorption determined by means of a short pre-exposure and the built-in technique charts of the system. All DR mammography units have at least one full automatic AEC mode, whilst mammography units using film-screen or CR technology may not have a fully automatic AEC. In semi-automatic mode, one or more technique factors are manually selected by the user, while the remaining factors are automatically determined; for instance, the radiographer chooses the A/F combination and the kV, and the AEC determines the appropriate amount of mAs, similar to how a phototimer would do. Such exposure mode was available on all the mammography units used with CRs, but has been suppressed on DR systems; the only manual adjustment still possible on some FFDM systems is the selection of the image detector area used as AEC during the exposure. This allows, in specific cases, to exclude some areas of the breast which would produce a peak of absorption during the pre-exposure, inducing the AEC to make wrong choices of the technique factors; this is the case, for example, for breasts with implants, where the goal of mammography is to investigate the tissues surrounding the implants. Finally, in manual exposure mode, the AEC is disabled and all the exposure parameters are manually selected by the operator; this is a modality always present on any mammography unit, typically used to expose test objects (or phantoms) during quality control and maintenance activities.

In Table 23.1, six examples of technique factors selected by the AECs of different digital mammography systems operating in full automatic are reported; they were obtained exposing three PMMA phantoms of 30, 45, and 60 mm thickness, respectively. It can be noted that there are significant differences among the

<table>
<thead>
<tr>
<th>TABLE 23.1</th>
<th>Technique Factors (A/F, kV, mAs) Selected by the Full Automatic AEC of Six Different Mammography Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>DR System</td>
<td>AEC Choice for 30 mm PMMA</td>
</tr>
<tr>
<td>Fuji Amulet</td>
<td>W/Rh</td>
</tr>
<tr>
<td>(a-Se FPD)</td>
<td>26 kV</td>
</tr>
<tr>
<td>61 mAs</td>
<td>97 mAs</td>
</tr>
<tr>
<td>GE Senographe DS</td>
<td>Mo/Rh</td>
</tr>
<tr>
<td>(Csl FPD)</td>
<td>27 kV</td>
</tr>
<tr>
<td>36 mAs</td>
<td>48 mAs</td>
</tr>
<tr>
<td>Dimension</td>
<td>W/Rh</td>
</tr>
<tr>
<td>26 kV</td>
<td>28 kV</td>
</tr>
<tr>
<td>54 mAs</td>
<td>91 mAs</td>
</tr>
<tr>
<td>IMS Giotto</td>
<td>W/Ag</td>
</tr>
<tr>
<td>25 kV</td>
<td>26 kV</td>
</tr>
<tr>
<td>96 mAs</td>
<td>161 mAs</td>
</tr>
<tr>
<td>Image 3DL</td>
<td>W/Al</td>
</tr>
<tr>
<td>32 kV</td>
<td>32 kV</td>
</tr>
<tr>
<td>11 mAs</td>
<td>16 mAs</td>
</tr>
<tr>
<td>Philips Microdose</td>
<td>W/Al</td>
</tr>
<tr>
<td>(photon counting)</td>
<td>32 kV</td>
</tr>
<tr>
<td>Siemens Inspiration</td>
<td>W/Rh</td>
</tr>
<tr>
<td>(a-Se FPD)</td>
<td>26 kV</td>
</tr>
<tr>
<td>60 mAs</td>
<td>110 mAs</td>
</tr>
</tbody>
</table>
parameters selected by the six systems for each phantom thickness, differences which depend on the characteristics of the X-ray source used by each manufacturer, the detector quantum efficiency, and the AEC calibration.

23.2.6 Image Post-Processing

As mentioned in the introduction, digital mammography allows the separation between the image acquisition process and the image display process (by means of dedicated monitors). This means that acquired images can be modified (post-processed), and processing can also be customized according to the individual preferences. In digital mammography, it is in general possible to handle two types of images: (1) “raw” and (2) “processed” images. Raw images are the acquired images after some technical adjustments such as bad pixel calibration and inhomogeneity corrections, where the signal per pixel is proportional to intensity and energy of interacting photons; they include a large amount of information (usually 14 bit images), but often linear lookup table makes them not suitable for human interpretation. Processed images are images derived from the raw images, modified to fit the response of the human eye (contrast equalization) and to show the whole breast area, from the pectoral muscle to the skin line (tissue equalization). Some manufacturers also apply soft edge enhancement algorithms, to improve spatial resolution, and/or local contrast enhancement algorithms, to improve local contrast in cases of breasts with significant glandular composition. In Figure 23.7, a raw image is compared to two processed images obtained by two different algorithms.

Another benefit of digital mammography is that the large amount of information captured by the raw images can be successfully used by computers to improve the performance. As an example, very complex software programs have been developed over time, with the aim of aiding radiologists in breast cancer detection, usually referred as computer-aided-detection (CAD) systems. Their development uses the progresses made in different disciplines of computer science, such as image processing, pattern recognition, applied statistics, and artificial intelligence (Doi 2015).

A CAD system has the goal of improving the detection capability of radiologists (see Section IV, Chapters 59 and 60 of this book). It is used as a “second opinion” after the radiologist has already evaluated mammography images; it might or might not suggest a potentially suspicious area, but the radiologist is asked to make the final decision, and might either agree with the computer and consider the CAD suggestion a real suspicion, or disagree and dismiss the CAD output. Its natural context is breast screening, a process where asymptomatic women in a certain age interval (typically 50–69 years) are invited to undergo mammography.

European breast cancer screening typically has radiographers taking the mammograms, followed by blinded double reading (two independent radiologists interpreting images without any knowledge about the other’s decision); women can be called back for further workup, in case a suspicious lesion is detected. In cases where only one of two readers decides for a recall, the woman is recalled anyway in some countries, while the decision for recall is postponed after the arbitration by a third reader. At recall, the woman gets additional mammography views, ultrasound, and so on, depending on the type of suspicious lesion. The double reading has been demonstrated to reduce the number of missed cancers compared to the single reading (Brown et al. 1996). The use of a CAD system was expected to be able to replace the second reader, while reducing screening costs. However, a systematic review by Azavedo et al. (2012) recently concluded that the scientific evidence is insufficient to determine whether the accuracy of single reading + CAD is at least equivalent to that obtained in standard

FIGURE 23.7 Raw versus processed images in digital mammography: (a) raw CC image of a dense breast acquired by a FFDM unit; (b) processed image obtained from image (a), applying a standard tissue equalization algorithm; (c) processed image obtained from image (a), applying a local contrast enhancement algorithm.
practice, that is, double reading, where two breast radiologists independently read the mammographic images.

Robert Nishikawa, one of the CAD pioneers, suggested, in a point/counterpoint paper published a few years ago in *Medical Physics*, that limitations in practical CAD applications result from human limitations in decision-making. In other words, despite correct CAD findings, radiologists may make the wrong decision (Nishikawa et al. 2006).

### 23.3 Radiation Dose in Digital Mammography

The parameter used to quantify the radiation dose received by the breast during mammography is the mean (or average) glandular dose (MGD or AGD) (see Section II, Chapter 29 of this book). It is defined as the average of all the absorbed doses to the different parts of the glandular tissue in the breast. The focus is on glandular tissue only, as this is considered the only breast tissue that is sensitive to radiation-induced cancers (detriment). The MGD can be obtained by measuring the entrance dose (incident air kerma), and multiplying it by one or more conversion factors, accounting for breast thickness and composition, and X-ray beam half-value layer (HVL). Conversion factors have been calculated through Monte Carlo techniques by different authors (Sobol and Wu 1997; Dance et al. 2000, 2009; Boone 2002). While breast thickness can be derived from the height of the compression paddle, breast composition (usually called “density”) can only be estimated. All simulations built to determine the dose conversion factors assumed that glandular tissue is distributed homogeneously in the breast (within a given contour of skin tissue). Conversion factors have been obtained for different glandular fractions. Patient specific dosimetry requires an assessment of this density. In population-based dose surveys, typical thickness related densities are assumed, based on detailed studies performed in the UK (Dance et al. 2000). Recently, it was reconfirmed that the typical breast composition is well below 50% glandular (Yaffe et al. 2009).

In terms of population dose, it was found that the mean MGD with digital mammography is around 1.4 mGy, versus a mean MGD of typically 1.9 mGy for screen-film mammography, that is, a 30% dose reduction of DR when compared with screen-film mammography (Gennaro and di Maggio 2006). Similar results (25% dose reduction) were published comparing dose from digital and screen-film mammography in the DMIST trial (Hendrick et al. 2010).

However, there are differences among digital techniques in mammography, in terms of both clinical performance and radiation dose. Weigel et al. (2014) have demonstrated that high performance results can be obtained in screening by photon counting systems, at radiation doses significantly lower than the doses delivered by other DR systems.

In contrast, with CR mammography, it is recommended to carefully verify the AEC setting of the mammography unit, avoiding its operating at low dose levels, with a potential negative impact on image quality. It has been estimated from screening performance results that, to provide the same diagnostic quality level as DR systems, CRs should work at a 60% higher dose (Bosmans et al. 2013).

### 23.4 Digital Breast Tomosynthesis

Clinical performance studies comparing digital to screen-film mammography succeeded to prove the non-inferiority of digital mammography, but not its superiority. The DMIST study was, thanks to its very large sample size, the first to be able to pick a subset of data, and more, in particular women with dense breasts and young women, for which digital mammography performed better than screen-film (Pisano et al. 2005). Recent multicenter trials have started to show more benefits from digital mammography in comparison to film-screen and CR technology (Séradour et al. 2014). The number of interval cancers keeps pointing to a need to improve the performance of breast imaging.

The reason why digital mammography cannot significantly improve the clinical performance is that both digital and screen-film mammography have the same inherent limitation, as any type of X-ray projection imaging. In fact, in mammography, any tissue between the X-ray beam entrance and the image detector is projected onto the image plane, resulting in an absorption map of overlapping structures. Tissue superimposition hampers image interpretation, either masking lesions and reducing their detectability, or creating false positive images. This is why mammography sensitivity (expressed in terms of capability in detecting cancers) dramatically falls as breast density increases (Moshina et al. 2016). Overlapping structures in projection images are also reported as “anatomical” or “structure” noise, as they constitute an obstacle to a correct diagnosis.

Digital breast tomosynthesis (DBT), made possible by digital detectors, is a quasi-3D technique which aims at solving the natural limitation of projection images by reducing the anatomical noise (see Section II, Chapter 20 of this book). A DBT clinical unit is very similar to a digital mammography clinical unit, or better, it is a digital mammography clinical unit with a few changes. In mammography, image acquisition always occurs with the X-ray source perpendicular to the image detector. Conversely, in tomosynthesis, the mammographic gantry is modified to allow X-ray source rotation, while the image detector is kept stationary. The breast is compressed as for the acquisition of a standard mammogram, the X-ray tube moves in an arc during the exposure, and a limited number of low dose projection images is acquired, at different angles over the arc. Tube motion can be either continuous (like for computed tomography) or step-and-shoot; in the latter case, each exposure is performed with the tube stationary at each angular position. If two structures are overlapped when imaged with the X-ray source perpendicular to the detector, they are shifted on the image plane when acquired with the tube angled. At the end of the acquisition of the full DBT projection series, a reconstruction algorithm is applied to reconstruct a stack of breast planes parallel to the detector plane. Those tomographic planes are often called “slices,” taking the term from Computed Tomography (CT) (see Section III, Chapter 32 of this book). The typical interval between adjacent tomographic planes is 0.5 to 1.0 mm. Their effective thickness is much larger. In Figures 23.8 and 23.9, the general principle of tomosynthesis acquisition and reconstruction are illustrated.

During the acquisition of the DBT projections, the position of the two objects embedded into the breast model (sphere and cube) at different depth (z-direction) is encoded by the shifts...
produced in each angled projection compared to their position in the 0° projection (obtained with the tube perpendicular to the image detector). For a certain tube angle, the deeper the object of interest (the closer to the detector), the smaller the shift produced, and vice-versa. The reconstruction algorithm reverses the acquisition process; by back-projecting the contents of the projection images, it derives the differences in object depth from their shifts. In the back-projected images (tomographic planes or slices), each object of interest is in-focus only in the tomographic plane corresponding to its physical depth into the breast. Any other object or structure not belonging to the same plane is out-of-focus, and appears as blurred in the tomographic plane. Thereby, the anatomical noise caused by structure overlapping is reduced in the reconstructed images, making clinical features easier to detect because of their higher conspicuity.

23.5 Clinical DBT Systems

Most of the clinical studies with DBT aimed to prove the added value of tomosynthesis in combination with mammography, rather than aiming to replace mammography with tomosynthesis. Population-based (screening) trials have demonstrated that the combination of tomosynthesis with digital mammography is superior to digital mammography alone, by significantly increasing the cancer detection rate (CDR), while reducing the recall rate (RR) (Ciatto et al. 2013; Skaane et al. 2013; Friedewald et al. 2014; Lång et al. 2015). Furthermore, in diagnostic applications, tomosynthesis is expected to be used as a complementary source of information, in addition to mammography.

There are several differences across the different technical solutions, regarding all the DBT components, as shown in Table 23.2. All the systems, with the exception of GE, use W anode target material, as available at the introduction of the first digital mammography units. Filter materials are chosen to produce more
penetrating X-ray spectra than those used for mammography, and the KV range used in DBT is higher (typically 30–40 kV) than the mammographic KV range (typically 25–32 kV). The image detector is mostly the same as used for digital mammography, but the read-out is tuned towards better performance for low dose projections. In addition, Hologic decided to increase the pixel size for DBT by rebinning, that is, by combining the content of two physical pixels (70 µm), in order to increase the SNR of the low dose projection images. Another exception is given by Fuji, that introduced a flat panel detector with hexagonal pixel shape (area equivalent to 68 µm²), permitting one to make the effective pixel area variable, changing the spatial resolution of the images. The choice about the tube rotation mode, either continuous or step-and-shoot, is equally distributed across the manufacturers. The continuous motion is faster, while the step-and-shoot motion is more accurate, but slower. The angular range, or sweep angle, is very variable among manufacturers, ranging between 15° (±7.5°) and 50° (±25°). The system by Fuji is designed to permit two different angular choices, 15° (standard mode) and 40° (high resolution mode). The number of projections is expected to be proportional to the angular aperture, but this is not systematically true. Three (Fuji, Hologic, and Planmed) out of six manufacturers acquire 15 projection images, two manufacturers (GE and IMS) acquire nine and 13 projections, despite the relatively wide angle, and, finally, the manufacturer (Siemens) using the widest angle, acquires 25 projections. All the systems, with the exception of IMS, equally distribute the tube current and system model. Iterative techniques iterate the image reconstruction a given number of times to better estimate these mathematical assumptions and generate images with lower noise. Because of the limited angles used in tomosynthesis compared to the full angular coverage of CT, image reconstruction in DBT is affected by different types of artifacts. A typical artifact is that produced by high contrast objects in the breast, like metallic clips, or large calcifications: the high contrast object is present, even shifted, in several planes adjacent to the in-focus plane. This artifact can be corrected by several algorithms (Wu et al. 2006). Another common artifact is the truncation artifact, due to the limited size of the image detector, which produces bright horizontal lines and a saturated area close to the detector edge. The truncation artifact can be minimized by reconstructing a volume a little larger than that corresponding to the compressed breast thickness, and applying different types of algorithms (Sechopoulos 2013).

### TABLE 23.2

Characteristics of the Six Commercial DBT Systems

<table>
<thead>
<tr>
<th>System</th>
<th>Fuji</th>
<th>GE</th>
<th>Hologic</th>
<th>IMS</th>
<th>Planmed</th>
<th>Siemens</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anode material</td>
<td>W</td>
<td>Mo or Rh</td>
<td>W</td>
<td>W</td>
<td>W</td>
<td>W</td>
</tr>
<tr>
<td>Filter material</td>
<td>Al or Rh</td>
<td>Rh</td>
<td>Ag</td>
<td>Ag</td>
<td>Ag</td>
<td>Rh</td>
</tr>
<tr>
<td>Detector</td>
<td>a-Se FPD</td>
<td>CsI FPD</td>
<td>a-Se FPD</td>
<td>a-Se FPD</td>
<td>a-Se FPD</td>
<td>a-Se FPD</td>
</tr>
<tr>
<td>Pixel (µm)</td>
<td>68-square</td>
<td>100</td>
<td>140b</td>
<td>85</td>
<td>85</td>
<td>85</td>
</tr>
<tr>
<td>Pixel shape</td>
<td>Hexagonal</td>
<td>Square</td>
<td>Square</td>
<td>Square</td>
<td>Square</td>
<td>Square</td>
</tr>
<tr>
<td>Tube motion</td>
<td>Continuous</td>
<td>Step-and-shoot</td>
<td>Continuous</td>
<td>Step-and-shoot</td>
<td>Step-and-shoot</td>
<td>Continuous</td>
</tr>
<tr>
<td>Angular range (°)</td>
<td>15 (STD)</td>
<td>25</td>
<td>15</td>
<td>40</td>
<td>30</td>
<td>50</td>
</tr>
<tr>
<td>No. projections</td>
<td>15</td>
<td>9</td>
<td>15</td>
<td>13</td>
<td>15</td>
<td>25</td>
</tr>
<tr>
<td>Dose (projection)</td>
<td>Uniform</td>
<td>Uniform</td>
<td>Uniform</td>
<td>Variable</td>
<td>Uniform</td>
<td>Uniform</td>
</tr>
<tr>
<td>Anti-scatter grid</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
<td>No</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>Reconstruction</td>
<td>FBPb</td>
<td>Iterative</td>
<td>FBPb</td>
<td>Iterative</td>
<td>Iterative</td>
<td>Iterative</td>
</tr>
</tbody>
</table>

a Rebinned from 70 µm pixel size.
b Filtered back-projection.

23.6 Reconstruction in DBT

As reported by Padole et al. (2015) conventional FBP is associated with a relatively high image noise and artifacts at reduced doses, because it is based on some mathematical assumptions, ignoring some key information, like statistics of X-ray photons and system model. Iterative techniques iterate the image reconstruction a given number of times to better estimate these mathematical assumptions and generate images with lower noise. Because of the limited angles used in tomosynthesis compared to the full angular coverage of CT, image reconstruction in DBT is affected by different types of artifacts. A typical artifact is that produced by high contrast objects in the breast, like metallic clips, or large calcifications: the high contrast object is present, even shifted, in several planes adjacent to the in-focus plane. This artifact can be corrected by several algorithms (Wu et al. 2006). Another common artifact is the truncation artifact, due to the limited size of the image detector, which produces bright horizontal lines and a saturated area close to the detector edge. The truncation artifact can be minimized by reconstructing a volume a little larger than that corresponding to the compressed breast thickness, and applying different types of algorithms (Sechopoulos 2013).

Several DBT “objects” can be reconstructed from tomosynthesis acquisitions: (1) Tomographic planes or slices, used as a stack of images parallel to the detector plane, and explored by scrolling or in a cine-loop mode. They permit one to assess the height of the lesion in the 3D volume. The number of planes depends on the compressed breast thickness and the sampling interval: for a 5 cm breast, the number of planes is 50 for a sampling interval of 1 mm, or 100 for a sampling interval of 0.5 mm. (2) Thick slabs: a given number of tomographic planes can be used together to build a sub-volume, whose thickness is usually about 1 cm. The purpose of slabs is twofold: on one side, allowing radiologists to scroll quickly through the breast; on the other side, to support detectability of groups of small clinical features present at several heights, like microcalcifications. (3) Synthetic mammograms: they are pseudo-mammographic
images, obtained from DBT reconstruction, “collapsing” all the tomographic planes onto the detector plane. The introduction of synthetic mammograms is recent (Skaane et al. 2014), and goes in the direction of replacing standard 2D mammography with tomosynthesis.

### 23.7 Radiation Dose in DBT

The mean glandular dose concept is also used to quantify radiation dose in tomosynthesis. New conversion factors for DBT X-ray spectra have been calculated, including a factor to account for the angular dependence of the X-ray paths (Dance et al. 2011; Sechopoulos et al. 2014). There are a couple of studies published about the comparison between radiation dose for one mammography view and one tomosynthesis view, using commercial systems. The first one is a phantom study, where MGD values delivered by a specific system (Hologic) were compared between FFDM and DBT using breast phantoms of different thickness and composition in automatic exposure mode. Considering phantoms with 50% glandular fraction (traditionally retained the “standard” breast composition), the thickness range is 2 to 8 cm, the MGD increase for a DBT view is between 0.8% for the 5 cm phantom and 76% for the 2 cm phantom; however, taking as a better approximation the typical breast composition, using the phantom with 14.3% glandular fraction, the MGD increase by DBT is between 67% and 135%, depending on the phantom thickness (Feig and Sechopoulos 2012). The second one is a small clinical study (270 patients) based on a different clinical system (GE), showing that MGDs for a single view DBT are comparable to digital mammography (Paulis et al. 2015).

However, the debate about radiation dose of DBT is not related to the MGD differences with mammography for a single view, but to the approach mostly followed up to now about the clinical use of digital breast tomosynthesis. As already mentioned, DBT has been proposed in clinical studies, in addition to standard mammography, with the side effect of increasing systematically the radiation dose of a factor depending on the total number of DBT views acquired. Despite the improved diagnostic accuracy proven by the combination of FFDM plus DBT versus FFDM alone, doubling radiation dose to the population is still an open issue. Nevertheless, the introduction of synthetic 2D images, obtained from tomosynthesis at zero-dose, aims to replace digital mammography with DBT. The non-inferiority of the combination of synthetic 2D plus DBT versus standard FFDM plus DBT has been demonstrated (Skaane et al. 2014). If the use of DBT alone, with the possibility of deriving synthetic 2D image, will become the standard in clinical practice, the subject about the increase in radiation dose caused by tomosynthesis, will be automatically closed.

### REFERENCES


