Chapter 7
Mammography

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7.1 Introduction

Statistics show that about one in eight women will develop invasive breast cancer in her lifetime in the United States. While there has been a decline in the mortality rate from breast cancer in recent years, it is still the second leading cause of cancer death in women according to the American Cancer Society (Siegel et al. 2014). The bright side is that breast cancer is one of the most treatable malignancies when detected early. If a patient's breast cancer is discovered and diagnosed in its early stages when tumors are small and local, the chance for successful treatment is close to 100%. Therefore, early detection of abnormal breast lesions is crucial for patient's long-term survival and thus reducing mortality (Swedish Organized Service Screening Evaluation, G 2006; Tabar et al. 2003; Tabar et al. 2011). Most cancer experts agree that among a variety of breast screening technologies, mammography is currently the most effective image modality for the early detection of breast cancer for its high sensitivity, excellent benefit to risk ratio, low cost, and low radiation exposure (Nyström et al. 2002; Tabar et al. 2003). Regular mammograms are recommended as a preventive measure for at-risk women and any woman aged over 40 in the states.

Mammography is a specialized radiographic examination of breast tissue using low-energy X-rays. It allows to identify anomalies (typically characteristic masses

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or microcalcifications) in the breast tissue that may be a sign of breast cancer. Mammography is primarily used as a screening and diagnostic tool for the early detection of breast cancer but also as a localization tool of suspicious area to guide needle biopsy and therapy. It can also be used to detect and evaluate breast changes and thus plays a role in treatment monitoring.

Albert Salomon, a German surgeon, used X-rays to examine mastectomy specimens in 1913. His study demonstrated the spread of breast carcinoma to the axillary lymph nodes and was considered the beginning of mammography. Stafford L. Warren, a radiologist at Rochester Memorial Hospital, New York published the first article on mammographic technique in 1930 (Warren 1930). Modern mammography started to advance in full swing in the late 1960s, when special X-ray machines were designed and used just for breast imaging. In 1965, radiologist Robert Egan at MD Anderson Hospital developed mammographic standards that became widely adopted in the field. The first mammogram machine was introduced in 1966, and dedicated mammographic equipment was used in France in 1974. By 1976 the mammogram had become the standard test for breast cancer detection. New technologies have been continually developed to reduce the amount of radiation required for a mammogram as well as to enable the detection of smaller lesions at an earlier stage. Since the Food and Drug Administration (FDA) approved the first digital mammography system in 2000, digital mammography has witnessed fast-pace advance in its technology. Digital mammography, unlike conventional mammography using film to capture and display the images, acquires the images using solid-state detectors and employs computer-aided viewing of the mammogram images, often resulting in a more detailed and accurate diagnosis.

In the early stage of X-ray mammography, direct exposure film was used without intensifying screen. Even though high-radiation doses were applied, the produced mammograms featured low contrast and thus poor diagnostic quality. As a matter of fact, they most likely did not provide much useful information for the early detection of breast cancer. In the late 1950s, Egan made significant improvement in mammographic imaging technology. He used a combination of a high milliamperage–low voltage technique, a fine-grain intensifying screen, and industrial film to produce mammographic images that were clearer and therefore easier to interpret. In 1971, motivated by dry processing technique developed by Xerox, mammography using the xeroradiographic process became very popular. While such mammograms showed good spatial resolution and sharpened edges at the cost of higher radiation doses, their contrast sensitivity remained poor. During the same period of time, technology in screen-film mammographic imaging systems was dramatically refined and improved. Screen-film mammography has been the mainstream in screening breast cancer since the early 1980s. Since its first public release in 2000, full-field digital mammography systems have shown great advantages in fast image acquisition and display with better image quality. Furthermore, anatomy-specific image processing and computer-aided detection tools have demonstrated to be capable of assisting the radiologist in identifying suspicious features in the images.

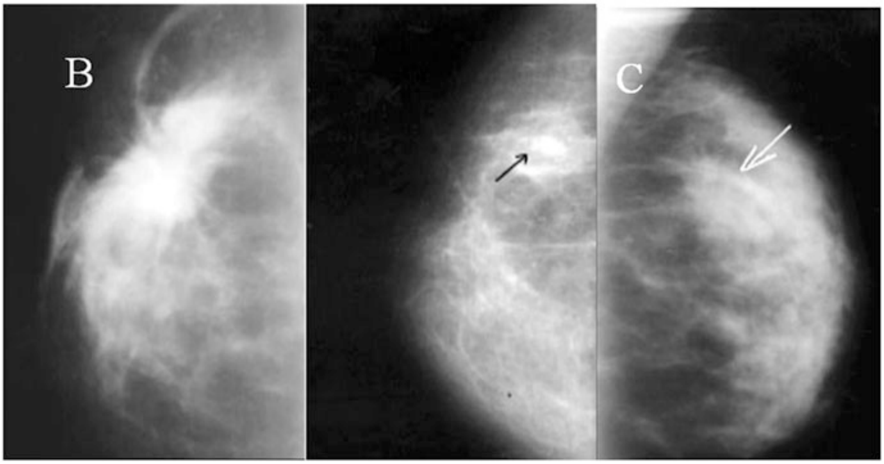


Fig. 7.1 Mammographic features of breast cancer. (**a**) mass with speculated margins, (**b**) clustered microcalcifications, (**c**) architectural distortion (Images are from Popli (2001))

Mammography can be classified into two main categories based on its functionality: screening and diagnostic mammography. Screening mammography attempts to look for signs of cancer in the asymptomatic women and thus is used as a preventative measure.

The primary mammographic signs of breast abnormalities include masses, clusters of calcifications, and architectural distortion (Fig.7.1). Masses with irregular or speculated margins raise high suspicion and need to be carefully evaluated. Calcifications are specks of calcium hydroxyapatite or phosphate. Their location, size, shape, and density are assessed to make the clinical decision. There is a large range of size for calcifications – from extremely small to several millimeters. It is important for screening to detect minute calcifications in breast tissues because of the high correlation between calcification patterns with disease. It has shown that mammography is an effective tool of early detection of breast cancer as it is able to detect calcifications as small as 100 microns. Architectural distortion is seen as straight lines radiating from a central area and retraction or bulging of a contour. Architectural distortion is often subtle and difficult to perceive without high-quality images.

The objective of screening mammography is to identify early-stage breast cancer when it is too small to be felt by palpation. Finding breast cancers early (before they have grown and spread) greatly improves the patient's chance for successful treatment. Mammography is the only validated imaging method in screening that contributes to reduction of mortality due to breast cancer (Swedish Organized Service Screening Evaluation, G 2006). In routine screening mammography, two X-ray images of each breast are acquired. The two images are usually in the mediolateral oblique (angled side view) and craniocaudal (head-to-foot) views.

More images may be necessary for large breasts in order to provide information on as much breast tissue as possible.

Another category is diagnostic mammography, which is performed to assist in the diagnosis of women with symptoms such as the presence of a lump. It can also be a follow-up examination when a suspicious area is identified during screening. Diagnostic mammography includes additional X-ray projections taken at different angles tailored to the specific area of concern. Furthermore, magnification views or spot compressions can be used to further assist the evaluation of the area. Magnification views amplify the mammographic image to show a specific area in greater detail. They are especially useful in evaluating microcalcifications. Spot compression applies compression to a local and smaller area of the breast, rather than the entire breast, achieving better separation of the breast tissue in the area of question. This technique facilitates easier visualization and assessment of suspicious areas. While diagnostic mammography alone is not capable of providing a definitive diagnosis of breast cancer, it can be used to evaluate whether breast abnormalities have a high likelihood of being malignant and whether a biopsy should be performed to confirm the existence of cancer.

There are both benefits and risks associated with any medical imaging procedures where ionizing radiation is utilized, and mammography is no exception. In general, the higher the radiation dose is applied to the organ (or area) in question, the better image quality (including resolution and contrast) can be obtained. However, unnecessary high radiation is hazardous to the patient as ionizing radiation may induce cancer. Since the breast is one of the most sensitive tissues to some adverse effects of ionizing radiation, a delicate balance between the mammogram quality and the radiation dose is demanded. The goal of mammography is to provide adequate information for appropriate medical decision-making, while keeping the radiation to the patient as low as possible. In the following sections, we will explain how current mammographic technologies achieve the trade-off between image quality and radiation dose.

7.2 Interactions Between X-Rays and Breast Tissues

Unlike general radiography, mammographic equipment operates at low X-ray tube voltages, typically 20–40 kV. The photon interactions in this energy range include the photoelectric effect and scattering processes. The photoelectric effect is the dominant interaction when the tube voltage is less than 22 keV. It causes most of the energy absorption from the incident X-rays and hence is the main source of the breast dose. There are two scattering processes in mammography: Rayleigh (also called coherent) scattering and Compton (also called incoherent) scattering. In Rayleigh scattering no energy is transferred between particles involved in the interaction, while there is transfer of energy between interacting particles in Compton scattering.

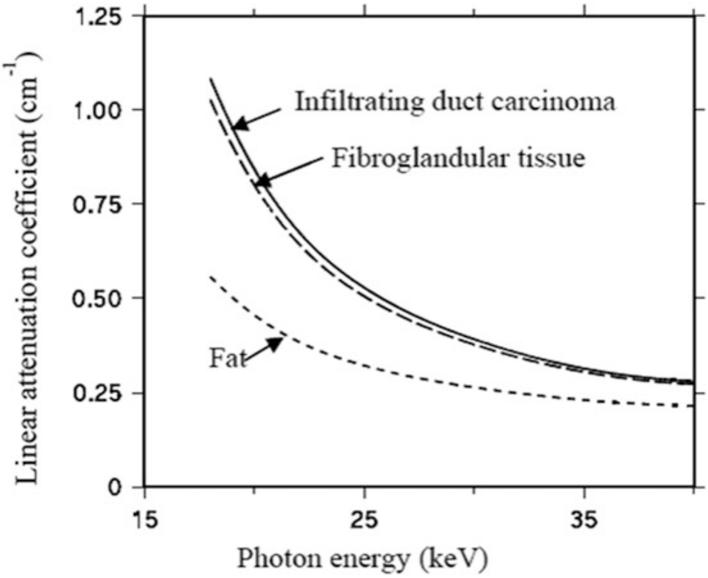


Fig. 7.2 Linear attenuation coefficient of different breast tissues as a function of X-ray energy. Small attenuation difference between the fibroglandular and cancerous tissues decreases with increasing X-ray energy (Data taken from Johns and Yaffe (1987))

Photon interactions between the incident X-rays and the breast tissues lead to the attenuation of the X-rays. The difference in composition of the normal and cancerous tissues in the breast results in different X-ray attenuation, which is used by mammography to accomplish screening and diagnosis. However, the attenuation differences (represented by the difference in linear attenuation coefficients) are very small. Figure 7.2 demonstrates the attenuating characteristics of three breast tissues: infiltrating duct carcinoma, fibroglandular tissue, and fatty tissue. Fibroglandular tissue has a much higher attenuation coefficient than fatty tissue. However, the difference in attenuation between fibroglandular tissue and carcinoma can be small. The attenuation differences between these tissues are decreased as the X-ray energy increases in the low-energy spectrum (<100 keV). The highest differences are observed at very low X-ray energies.

Another important factor of image quality for breast imaging is subject contrast. Adequate subject contrast is required for the detection of the minute difference between the normal and cancerous breast tissues. Figure 7.3 shows that the contrast of the ductal carcinoma declines with energy. Both the differential attenuation and subject contrast characteristics between the normal and malignant tissues require that mammography operates at a low energy level for the best screening and diagnostic capabilities.

As mentioned earlier, photoelectric effect occurs with high probability in the low X-ray energy range. Heavy photoelectric effect at very low energy range significantly increases the absorption dose and exposure time for the patient. Thus there must be a compromise between image quality and patient dose for mammography.

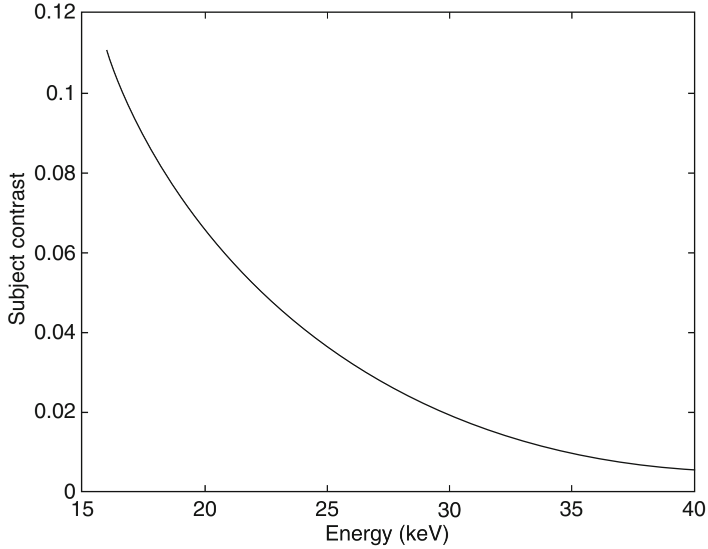
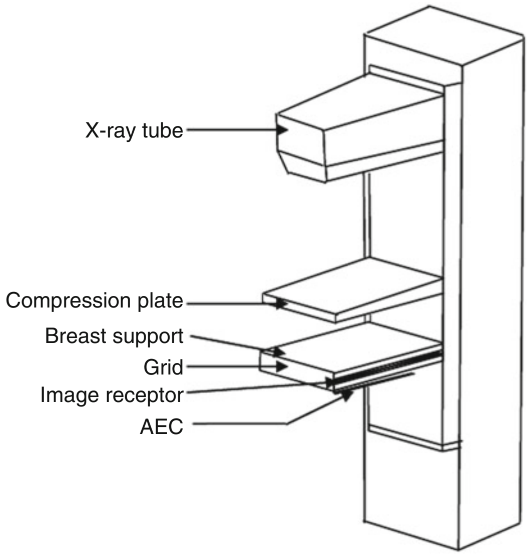


Fig. 7.3 Rapid declining of contrast of the ductal carcinoma indicates that it is necessary to use a low-energy X-ray tube for mammogram to successfully detect breast cancer (Adapted from Yaffe (1995) and Haus and Yaffe (1994))

Fig. 7.4 Components of a typical mammography system



In order to minimize patient dose while acquiring high-quality mammograms for optimal detection of breast cancer, mammography employs a complex system of technology including dedicated X-ray equipment, specialized X-ray tubes, compression devices, anti-scatter grids, and sensitive image receptors (refer to Fig. 7.4 for an illustration). Modern mammography also utilizes sophisticated image processing and viewing components to maximize the likelihood of detecting small breast tumors. Computer-aided detection and diagnosis tools can further improve the performance of mammography.

7.3 Imaging System

The mammogram is produced when X-rays irradiate the breast, and the transmitted X-rays are recorded by an image receptor. The mammogram exhibits the differential attenuation of X-rays along paths through the structures of the breast.

Currently, there are two distinct mammographic techniques primarily based on image formation: analog and digital mammography. Analog mammography refers to screen-film mammography. X-rays are captured using a screen-film cassette, and the image is recorded and stored on the film. The film is subsequently hung on a viewing board for the reading radiologist.

Digital mammography is a rapidly progressing newer technique. In digital mammography, X-ray beams are captured on a specially designed digital detector. This detector then converts the X-ray photons into electronic signals, which are then transferred to a computer. Digital mammography can reduce 30–40% radiation dose. The computerized image is then available for the radiologist to review on a high-resolution monitor. Images may be manipulated by the radiologist using the tools such as magnifying, masking of light, windowing, leveling, and comparison to prior mammograms on the computer.

In spite of the different image acquisition and display principles used in analog and digital systems, they share some components in the procedure of mammographic imaging, especially before the X-ray beams strike on the image receptors. We will discuss the common components without specifying analog or digital systems while devoting some sections to their individual components.

The goal of mammography is to achieve the image quality required for a given detection task while keeping the radiation dose absorbed by the patient as low as reasonably achievable (ALARA principle) (Huda et al. 2003). To achieve the optimal balance between high-image quality and low radiation exposure, the mammographic unit is specifically designed for examination of the breast tissues. The main components of a typical mammography system are illustrated in Fig. 7.4. The patient may be examined standing or sitting, with her breast resting on a support plate. The X-ray tube and breast support plate are mounted on a mechanical assembly. The assembly may be rotated to achieve different projection angles. Its height can be adjusted to accommodate patients of different size. An anti-scatter grid is placed between the breast support and image receptor. An image receptor is a device used to absorb X-rays transmitted by the breast and acquire the image. The receptor is a screen-film cassette for traditional screen-film mammography and digital detectors for digital mammography. Firm compression is applied to the breast using a plastic compression plate. An automatic exposure control (AEC) device is used to adaptively adjust the exposure amount to the individual patient. It terminates the radiation when the required clinical quality is achieved in the image or the permissible amount of radiation dose is exceeded. All these components have been optimized to deliver the best trade-off between the mammogram quality and radiation exposure.

7.3.1 X-Ray Tube

Low-energy X-rays for mammographic imaging are produced in a specially designed tube housed in a metal envelope. The window of the X-ray tube is made of beryllium (not glass) with a maximum thickness of 1 mm. Electrons are emitted from a heated negatively charged cathode. They are accelerated in an imposed electric field and focused to strike a positively charged anode. The area on the anode upon which the X-rays impinge is referred to as the *target* or *focal spot*. The X-ray tube is specially designed and constructed for imaging the soft tissue of the breast. The range of X-ray tube operating potentials is from 20 kV to 40 kV for clinical imaging. The exact value of the tube potential is chosen based on the thickness and composition of the breast for a specific patient.

7.3.1.1 Anode

Mammographic X-ray tubes use rotating anodes. The most commonly used material for anode is molybdenum (Mo, $Z = 42$). Rhodium (Rh, $Z = 45$) and tungsten (W, $Z = 74$) are also used. Molybdenum and rhodium are chosen for the target material primarily because of their desired X-ray spectra. The spectra consist of both bremsstrahlung radiation and characteristic X-rays specific to the target material. Characteristic X-rays are of particular interest in mammographic imaging. Characteristic radiation occurs at 17.5 and 19.6 keV for molybdenum and 20.2 and 22.7 keV for rhodium. These characteristic X-ray energies satisfy the requirement of mammography's low-energy spectrum (~20 keV) to provide adequate discrimination between cancerous and normal breast tissues (referred to Fig. 7.2).

The X-ray spectrum from a molybdenum target at 25 kV is shown in Fig. 7.5a. The low-energy bremsstrahlung X-rays deliver significant breast dose with little contribution to the clinical capability of the image. The higher-energy bremsstrahlung X-rays significantly reduce subject contrast. To mitigate undesired low- and high-energy bremsstrahlung X-rays, additional filters are needed for the target. These filters are often made of the same material as the target so that they can reduce the low- and high-energy bremsstrahlung X-rays but allow efficient transmission of the characteristic X-rays. Usually a filter of molybdenum is used with a molybdenum target (Mo/Mo target/filter), and a filter of rhodium is used with a rhodium target (Rh/Rh target/filter). An exception is that a molybdenum target can be combined with a rhodium filter (Mo/Rh target/filter), which performs exceptionally well for imaging thicker and denser breast. This combination produces a

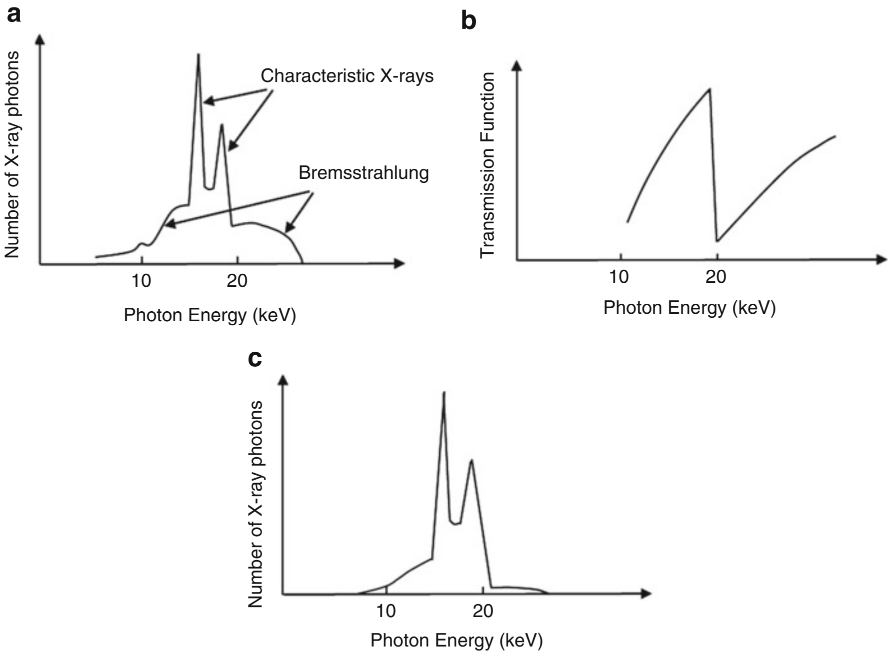


Fig. 7.5 (**a**) X-ray energy spectra for a molybdenum anode, including both bremsstrahlung and characteristic X-rays; (**b**) transmission function of a molybdenum filter of 30 μm thick. The molybdenum K-edge is at ~20 keV, resulting in reduced X-ray transmission above the K-edge; (**c**) X-ray energy spectra for a molybdenum anode with a molybdenum filter of 30 μm thick

slightly higher effective energy than the Mo/Mo target/filter, allowing transmission of X-ray photons between 20 and 23 keV.

The transmission property of a 30 μm thick molybdenum filter is shown in Fig. 7.5b. This filter attenuates both X-rays in the low energy range and those above its own K-absorption edge, while the characteristic X-rays from the molybdenum target pass through the filter with high efficiency. This type of filter is called K-edge filter.

The X-ray spectrum from a molybdenum target at 25 kV with this filter is shown in Fig. 7.5c. Comparison of targets without (Fig. 7.5a) and with filters (Fig. 7.5c) shows the significant removal of undesired X-rays and relative enrichment of X-rays in the range of 17–20 keV, ensuring the Mo/Mo target/filter combination suitable for mammographic imaging.

Since the atomic number of rhodium is higher than that of molybdenum, the X-ray spectrum using a rhodium target is harder than that for a molybdenum target. Thus rhodium anodes offer advantages for imaging the thicker, denser breast at a lower absorbed dose than with the molybdenum anodes. Multiple targets or filters are commonly included in a modern X-ray tube design. The technologist selects the appropriate target/filter combination based on the size and density of the breast to be examined.

7.3.1.2 Focal Spot Size

High resolution and high contrast are essential for mammographic images in order to visualize small calcifications and fine structures in the breast. The size of the focal spot of the X-ray tube is one of the main factors that determine the image resolution. Most X-ray tubes used for mammography have small focal spots, typically 0.3–0.4 mm. These are less than half the size used in general radiography. Such small focal spots are needed to image fine detail such as microcalcifications, whose size may be less than 100 μm. For general mammography purposes, a dual-focus X-ray tube is usually required. An even finer focus spot (0.1–0.15 mm) is used for magnification techniques exclusively. The tube current is 75–125 mA for the large focal spot (0.3 mm) and 15–35 mA for the small focal spot (0.1 mm).

7.3.2 Breast Compression

Since mammography is projection imaging, the highly irregular and easily varying structures in the breast overlap one another in the mammogram. The superimposition of different structures often causes difficulty in distinguishing tissue features. Compressing the breast spreads apart the structures and hence allows better visualization of the breast tissues. When the breast is compressed thinner, less X-rays are needed to penetrate the tissue, and thus the radiation dose will be lower to achieve images of similar quality. Furthermore, compressing the breast reduces the thickness of the back of the breast (close to the chest wall) and makes the breast under examination more uniform in thickness. Hence, tissues near the chest wall are less likely to be underexposed, and tissues near the nipple are less likely to be overexposed. An immediate benefit of minimizing over- and underexposure in regional areas of the breast to X-ray radiation is that the mammographic image is easier to interpret.

Making the breast thinner with more uniform thickness, breast compression reduces scattered radiation and beam hardening amount that occur as the X-ray beam passes through the tissues and thus improves the image contrast. Motion of the breast can greatly blur the image and thus make it impossible to observe small details. Breast compression will hold the breast still so motion is restricted. In summary, breast compression is essential in mammography for immobilizing the breast, separating superimposed tissue components, reducing scattered radiation, reducing radiation dose, and facilitating image interpretation (Saunders and Samei 2008). All dedicated mammographic systems have a built-in stiff compression device that is parallel to the image receptor.

7.3.3 Anti-scatter Grids

X-rays transmitted through the breast consist of primary and scattered radiation. Primary radiation is the radiation that passes through the breast without absorbed or scattered by the breast tissue. It contains the information regarding the attenuation characteristics of the breast. On the other hand, scattered radiation recorded by the image receptor does not reflect the characteristics of the breast tissue. Instead, it is an additive radiation that generates a noisy background to the image. Therefore scattered radiation can significantly degrade contrast of the breast tissue of interest in the mammographic image. The degradation of contrast depends on photon energy, breast size, and image receptor characteristics. It can be quantified using the contrast degradation factor (CDF) as defined by

$$\mathrm{CDF} = \frac{\text{Image contrast with scattered radiaiton}}{\text{Image contrast without scattered radiaiton}}$$

The CDF decreases with thicker breast. For example, the CDF for a 50 mm thick breast is 0.65, resulting in a very low quality of the mammographic image. Low-quality mammographic images are inadequate for the detection of subtle features indicating malignant tissue. This urges adoption of anti-scatter techniques to improving contrast. In mammography, reduction of scattered radiation is commonly achieved by utilizing anti-scatter grids.

X-ray photons transmitted through the breast and the breast support platform are incident on an anti-scatter grid. An illustration of an anti-scatter grid can be found in Fig. 7.6. The primary beams pass through the grid, while scattered radiation is maximally attenuated by the grid. There are two types of anti-scatter grids: stationary and moving grids. The stationary grid makes use of high line density, for instance, 80 lines/cm grid to prevent scattered X-rays from reaching the image receptors. The interspace material is usually aluminum. The moving grid employs

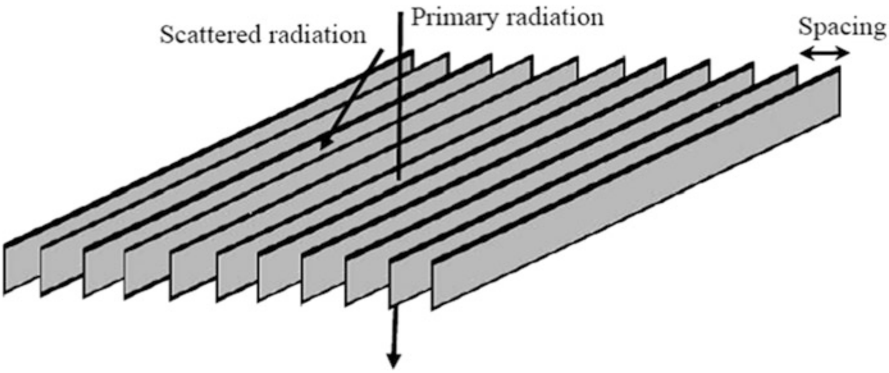


Fig. 7.6 Illustration of an anti-scatter grid

Table 7.1 CIF and BF values for the Philips moving anti-scatter grid

Breast thickness (cm)	CIF	BF
2	1.25	1.68
4	1.38	1.85
6	1.54	2.06
8	1.68	2.24

lower line density, e.g., 30 lines/cm, and paper or cotton fiber is used as interspace material.

Anti-scatter grids do not only reduce the amount of scattered radiation, but they also attenuate primary radiation. A higher radiation dose is needed in order to compensate for such unintended attenuation of primary radiation.

The performance of the anti-scatter grid can be assessed in terms of the contrast improvement (CIF) and Bucky factors (BF). The CIF is the ratio of the contrast with the grid to that without the grid. The BF represents the increase in radiation dose associated with the use of grid. The values for CIF and BF for the moving Philips anti-scatter grid are shown in Table 7.1. Significant improvement in contrast is achieved in thick breast (e.g., 8 cm thick breast) at the cost of increased breast dose.

After most scattered photons are removed by the anti-scatter grid, a large portion of the remaining X-rays are primary radiation and reflect the breast tissue characteristics. The remaining X-rays will impinge on the image receptors to form the image.

7.3.4 Mammography Image Receptors

Image receptors are devices that are used to absorb X-rays transmitted by the breast (after anti-scatter grids) and to acquire the mammographic images. When the X-rays strike on the image receptor, they interact and deposit most of their energy locally. Two types of image receptors will be discussed here: screen-film combination for conventional mammographic systems and digital detectors for digital mammographic systems.

7.3.4.1 Screen-Film Mammographic Image Receptor

When first introduced, mammographic systems used direct exposure radiographic film to obtain the high spatial resolution required at the cost of high radiation dose. Since the mid-1970s, intensifying screens were used in conjunction with radiographic film and greatly reduced the radiation exposure while maintaining the high image resolution. At present, the most common image receptor used in conventional mammography is the screen-film combination (shown in Fig. 7.7). The receptor comprises a film mounted onto an intensifying screen. The screen-film combination is housed in a light-tight cassette. Intensifying screens are used to

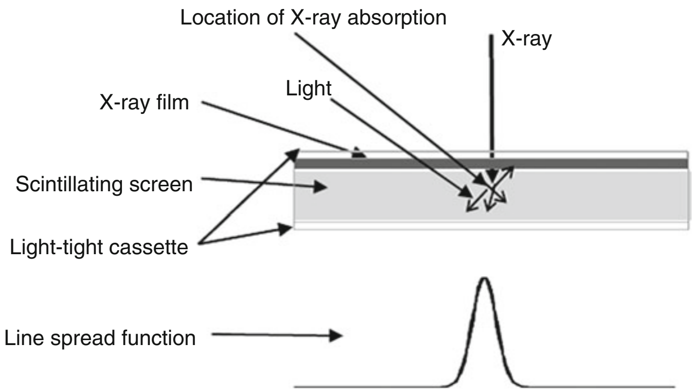


Fig. 7.7 Illustration of the components and blurring effect of screen-film mammographic image receptors

capture X-rays and then emit visible light. When an X-ray is absorbed by the intensifying screen, the light scintillation produces several light photons. The film next to the screen captures these light photons and produces the mammographic image.

Intensifying screens are highly effective in this type of image receptor because film is much more sensitive to light than to X-radiation. Without intensifying screens, approximately 100 times as much X-radiation would be required to expose a film of similar quality. The major drawback of using intensifying screens is that they introduce blurring into the imaging process, which is quantified by a line spread function (illustrated in Fig. 7.7). The blurring effect lowers the image resolution and needs to be carefully considered when selecting screens for specific clinical applications.

The thickness of the intensifying screen shapes the spreading of the line spread function and thus determines the quality of the mammographic image. Thicker screens are more dose efficient as they capture more X-rays. In the meantime, thicker screens also cause more light scatter, and their line spread functions become wider. This causes blurring and thus lowers resolution of the image. Appropriate screen thickness must be selected to compromise radiation dose and image quality based on the specific clinical application.

Since the intensity of transmitted (and thus attenuated) X-rays is inversely proportional to an exponential function of the transmitting distance, the film should be placed as close as possible to the X-ray source to maximize the intensity of the X-rays incident onto the film. By doing this, the resultant mammographic image shows the best quality given the same amount of exposure to the breast. In a screen-film mammography system, the film is placed next to the surface where the X-rays enter the scintillating screen.

After decades of steady research and technical advancement, current screen-film systems produce mammographic images with high spatial resolution, which is

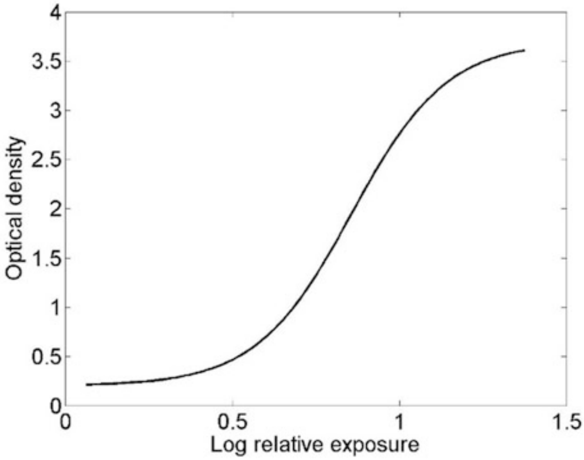


Fig. 7.8 Film characteristic curve for a screen-film mammographic system

suited for detection of fine structure such as microcalcifications of a considerably small size. However, there are inherent limitations to further technical improvement for such systems.

The performance of the screen-film receptor is strongly limited by the characteristics of the film. Although the conversion of X-ray to light photons by the intensifying screen is almost a linear transformation, the response of the film to light photons is nonlinear as shown in Fig. 7.8. The achievable contrast of the film is proportional to the gradient of this curve. The gradient of the central section of the response curve is steep, and over this exposure range, small differences in contrast can be detected on the developed film. Unfortunately the response curve is very flat at both high and low exposures, which results in very little change of optical density seen on the processed film over largely varying X-ray exposure. Under such circumstances it is very difficult to distinguish different tissues over these exposure ranges. The range of X-ray exposure over which the film response gradient is adequately large for clinical imaging is called *dynamic range*. Figure 7.8 shows the narrow dynamic range for the film response. Such narrow dynamic range of the film requires strict exposure conditions. Otherwise the resultant image can be of poor image quality, and breast imaging needs to be repeated, especially when imaging low-contrast lesions in dense breasts (Pisano et al. 2000).

Another major problem is film granularity, which introduces structural noise into the image and thus reduces visibility of microcalcifications and other fine details within the breast. Furthermore, a large amount of storage space is needed for mammographic films. As mammogram became the standard screening test for breast cancer, most women have their mammogram performed every 1 or 2 years. The demand for film storage grows dramatically. Film also must be physically transported to the physician for viewing, causing extra labor and time delay.

7.3.4.2 Digital Mammographic Detectors

Digital detectors are the image receptors for digital mammographic systems. They were developed to overcome the limitations of screen-film systems mentioned above. In digital mammographic systems, images are generated and stored as a digital signal. Transfer and storage of such images are achieved using electronic devices (including computers and networks), and thus no hard copy storage and distribution are needed as by film, saving both time and space.

Unlike stringent requirements for proper exposure demanded by film receptor, digital detectors provide a much larger dynamic range of operation, improving visualization of all areas of the breast. One of the most pronounced advantages of the digital system is the involvement of advanced computer processing of the acquired mammographic images. Stored in digital format, the image content can be manipulated to optimize contrast for each individual clinical task, accomplished by sophisticated and powerful image processing techniques. Computer-aided diagnosis further assists physicians in the interpretation of mammographic images to derive earlier and more accurate diagnostic outcome.

Two methods of image capture have been used in digital mammography, representing different generations of technology: indirect conversion and direct conversion.

Indirect Conversion Digital Detectors

Indirect conversion detectors were used in early digital mammography systems in the states. The GE 2000D or Fischer SenoScan uses indirect flat panel detectors made with cesium iodide (CsI). Such detectors use a two-step process for X-ray detection and image formation, as shown in Fig. 7.9. Similar to the screen-film system, a scintillating layer is first employed to capture the X-ray photons and convert them to light. The difference and advancement of digital detectors over the film-screen combination is shown in the next step. Instead of the film used in the screen-film system, an array of thin-film diodes (photodiodes) converts light

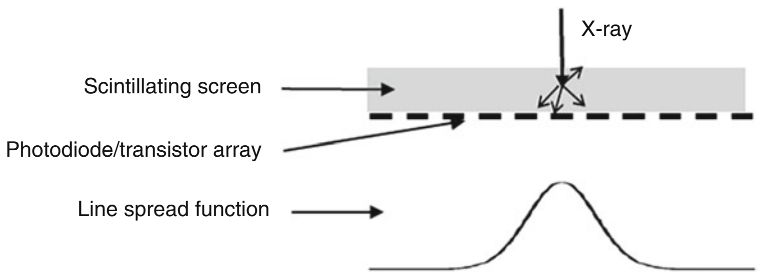


Fig. 7.9 Indirect conversion detectors utilize a scintillating layer to absorb the X-ray and to generate light photons. A photodiode array detects the light photons, and a transistor array converts the light photons to electronic signals

photons to electronic signals. Electronic signals are subsequently captured using thin-film transistors, and the digital mammographic image is produced, i.e., film imaging is replaced by digital imaging using thin-film diodes/transistors. We can say that digital mammography using indirect conversion detectors is a digital extension of screen-film imaging to some extent. Similarly to screen film, a performance compromise between the image resolution and radiation dose has to be made for indirect conversion digital mammography.

Despite the great performance similarity between the indirect conversion digital mammography and screen-film mammography, the digital evolution of the film receptor to the indirect conversion digital detector gives rise to complications for the placement of the scintillator. In screen-film systems, film is placed next to the surface where the X-rays enter the scintillator in order to achieve the highest intensity for the mammographic image. In the indirect conversion digital system, because X-rays cannot pass through the photodiode/transistor array, the array must be placed next to the scintillator surface that is farthest from the X-ray source. With this setup additional scattering of photons through the scintillator results in a wider line spread function, lowering the spatial resolution compared to the screen-film system.

Direct Conversion Digital Detectors

Direct conversion digital detectors are also called photoconductors. Systems like the Hologic Selenia or Siemens Novation use direct flat panel detectors made with amorphous selenium (α-Se). The layer of α-Se in the detector absorbs the X-ray photons transmitted by the breast and directly converts them to an electronic signal that is linearly proportional to the intensity of the photons (illustrated in Fig. 7.10).

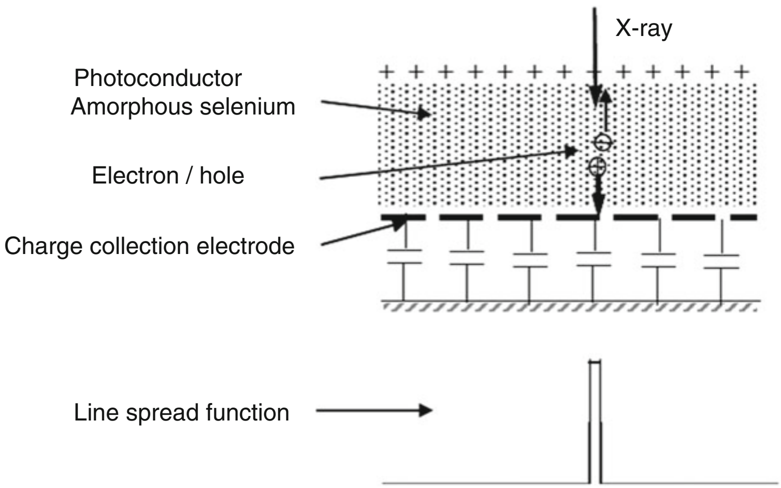


Fig. 7.10 A direct conversion detector uses a photoconductor to absorb the X-ray and directly generate the signal

These direct conversion digital detectors completely replace the screen-film combination, and there is no need for the intermediate step of light emission resulted from the scintillator. An external electric field is imposed across the detector system. As the photoconductor absorbs the incident X-rays, it produces electron hole pairs (see Fig. 7.10). The imposed electric field drives the electron (hole) to the photoconductor's surface with the positive (negative or ground) potential. As illustrated in Fig. 7.10, the holes drift toward a charge collection electrode and are then collected on a capacitor. Because the electrons and holes travel along the direction of the external electric field, there is no laterally spreading charge. This results in an exceptionally narrow point spread response, of about 1 micron (illustrated in Fig. 7.10). The immediate consequence of such narrow line spread functions is superior spatial resolution in digital mammographic images.

In direct conversion detectors, the sharpness of the response function is independent of the thickness of the photoconductor. Thicker photoconductors do not compromise image resolution; however, they attenuate much more X-rays. In practice, sufficiently thick photoconductors are commonly used in order to stop the majority of the incident X-rays without adversely affecting the spatial resolution. Therefore, a significant advantage of direct conversion digital detectors is that they can simultaneously achieve both high resolution and low radiation exposure. Based on the efficiency of direct conversion and its elimination of light scatter, direct conversion digital detectors are able to offer higher image resolution compared to the indirect conversion detectors. The major drawback of direct conversion systems is that they are more costly.

7.3.5 Automatic Exposure Control System

Breasts vary in composition and thickness, so the imaging parameters (X-ray tube voltage, tube current, scan time, etc.) for mammography should be adaptively selected for individual patients to optimize image quality and reduce radiation dose. For example, the time duration of radiation exposure needed to achieve desirable diagnostic quality mammograms vary with different sizes and densities of the breasts. It is difficult for the technologist to estimate the attenuation caused by the breast by visual inspection, and therefore all mammography units are equipped with automatic exposure control (AEC). AEC enables consistently optimal image exposure despite variations in tissue density, thickness, and the user skill level.

In screen-film mammographic systems, the AEC sensor is placed behind the film cassette in the imaging flow (see Fig. 7.4). If the AEC sensor were placed in front of the imaging cassette as in conventional radiography, it would attenuate the X-rays too severely and cast its own X-ray shadow on the mammographic image. The shadow is very severe especially at the low X-ray energies used in mammography. The sensor is connected electrically to an exposure control circuit. When the AEC system has detected the predetermined amount of radiation transmitted through both the breast and the cassette, the circuit automatically terminates the exposure. The location of the sensor must be adjustable so that it can be placed beneath the appropriate region of the breast in order to obtain adequate image signal in that region.

Due to the complexity of soft tissue in the breast, the attenuation of X-rays is highly heterogeneous over the whole region of the breast. The signal from the AEC sensor is heavily affected by such regional inhomogeneity in attenuation of the breast tissue. Therefore, the size of the sensor and its location under the breast can greatly affect the exposure used to acquire the mammogram. In modern mammographic equipment, AEC is computer controlled so that sophisticated automatic corrections can be made during the exposure (Haus and Jaskulski 1997). For screen-film mammography, AEC devices play a critical role to maintain a desired optical density on the processed film, independent of variations in breast attenuation caused by spatial variations in tissue composition and thickness, X-ray tube voltage setting, or imaging field size.

Full-field digital mammography systems cannot employ conventional AEC methods because digital receptors have absorbed almost all the X-ray beam. As a matter of fact, the design of AEC in the digital system is simpler. The digital detector itself is used as the AEC sensor. Since the image brightness and contrast can be easily adjusted using the computer display, the goal of including AEC device in the digital system is different from that for the screen-film system. Here, the exposure level is set to achieve a desired image signal-to-noise ratio. With the assistance from AEC, direct conversion digital detectors optimize the conversion from the X-ray radiation to electronic signals and thus keep image quality approximately constant while maintaining patient dose as low as reasonably achievable (Huda et al. 2003).

Because of film's narrow exposure dynamic range, appropriate AEC operation in screen-film systems is critical. On the other hand, digital detectors have a much larger dynamic range and consequently are more tolerant of exposure variations. Therefore, in digital mammography, image retaking occurs less frequently, and thus patient dose can be potentially reduced compared to film-screen mammography. Digital mammographic systems also employ more advanced AEC methods with the assistance of computer technologies. The function of an AEC can be expanded to automatically choose tube voltage and current. The digital system can be designed to obtain a fast low-dose pre-image. AEC will analyze the pre-image in real time and determine whether the selected tube voltage would achieve the desired image quality in a short scan time. If not, an optimum tube voltage–current combination will be selected to ensure that the exposure time limit is not exceeded. Information on breast density can also be measured and used to further improve the functionality of the AEC system.

7.4 Digital Mammography

With the technical perfection of mammographic components including the X-tube, focal spot size, breast compression, anti-scatter grid, and screen-film cassette, screen-film mammography is capable of producing images with high spatial resolution and contrast at a low-dose radiation. It has been the gold standard in screening breast cancer. However, there exist limitations in its ability to acquire and display the finest or most subtle details and to produce images at the most efficient radiation dose to the patient.

With the extraordinary advance of computer technology in everyday life, the transfer of imaging from film to the digital format started three decades ago with the introduction of digital radiography. However, the transition from conventional mammography to its digital counterpart was not intermediate because it was challenging to design and develop a full-field digital detector (Van Ongeval et al. 2006). The first full-field digital mammography unit was approved for sale by the Food and Drug Administration in January 2000 (Pisano et al. 2004). Since then a large number of hospitals and medical centers worldwide have installed and screened patients with digital mammographic systems.

The most outstanding difference between digital mammography and conventional screen-film mammography lies in the fact that image acquisition is decoupled from image display, archiving, and retrieval in digital mammography (Feig and Yaffe 1995; Pisano et al. 2000). In conventional mammography, the image is obtained and displayed on the same film, and thus the functionalities of image acquisition and display are highly correlated through the same media–film. In digital mammography, the image is acquired as an electronic signal by the digital detectors which subsequently stored in a computer. The digital image can later be displayed either in the format of "soft copy" or "hard copy." In soft copy display method, the image is viewed on a high-resolution video monitor. In hard copy display method, the image is printed onto a light-sensitive material such as film. The film is then viewed on a light box as in the conventional screen-film system.

The independent operation of image acquisition, display, and storage in digital mammography allows for optimization of each process separately. For example, acquisition is performed using highly efficient, low-noise X-ray digital detectors. Stored digitally, the image can be displayed on the monitor with adjusted contrast based on the radiologist's diagnostic criteria. Image display is completely independent of the detector's properties and solely relies on the monitor's performance characteristics and the computer's capability of manipulating image gray scales. Digital mammograms can be stored on a variety of storage devices, including hard disk drives, optical disks, etc., which is much less bulky compared to the screen-film system. Again, the storage has nothing to do with either the digital detectors or display monitors.

The complete decoupling and full optimization of the digital detector, the viewing system, and the storage device in the digital system leads to impressive advantages over the screen-film systems. The digital mammography offers a larger dynamic range, remarkably improved contrast and considerably increased signal-to-noise ratio. As a result, digital mammography exhibits increased sensitivity and specificity in breast cancer detection.

One of the most valuable benefits of digital mammography is that because the image data are presented in digital form, advanced data processing techniques can be developed to fully utilize the data quantitatively in specialized applications. Data (image) processing has been widely used to enhance the acquired digital mammographic images. Any beneficial processing techniques, for instance, contrast enhancement, edge sharpening, and noise reduction filtering, can be conveniently applied to improve the image appearance. Another class of image processing techniques can be used to analyze the mammogram with feature identification tools to search for signs of cancer. This approach is called computer-aided diagnosis (CAD) and provides invaluable information and assistance to the physician for better and faster diagnosis. Other advanced applications made possible through digital imaging, such as dual energy and 3D tomosynthesis, are expected to further improve diagnostic sensitivity and specificity.

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