CHAPTER 8

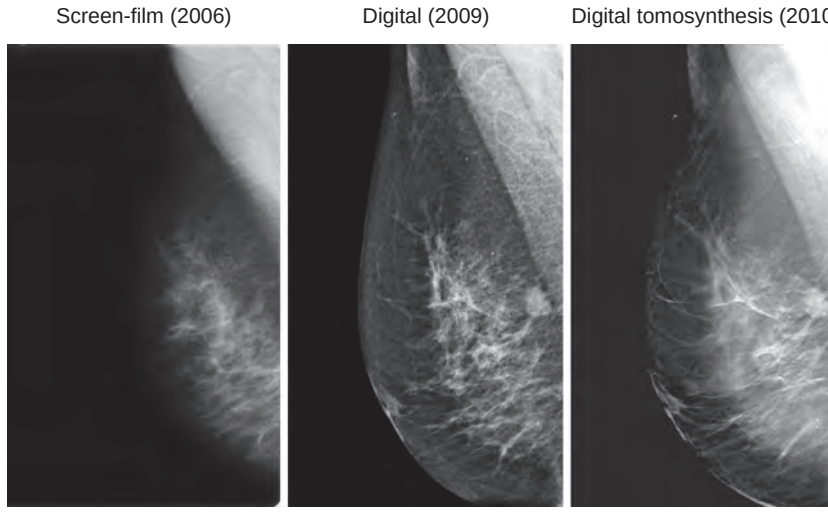
Mammography

Mammography is a radiographic examination that is designed for detecting breast pathology, particularly breast cancer. Breast cancer screening with mammography assists in detecting cancers at an earlier, more treatable stage, and is an important clinical procedure because approximately one in eight women will develop breast cancer over their lifetimes. Technologic advances over the last several decades have greatly improved the diagnostic sensitivity of mammography. Early x-ray mammography was performed with direct exposure film (intensifying screens were not used), required high radiation doses, and produced images of low contrast and poor diagnostic quality. Mammography using the xeroradiographic process was very popular in the 1970s and early 1980s, spurred by high spatial resolution and edge-enhanced images; however, its relatively poor sensitivity for breast masses and higher radiation dose compared to screen-film mammography led to its demise in the late 1980s. Continuing refinements in screen-film technology and digital mammography, which entered the clinical arena in the early 2000s, further improved mammography (Fig. 8-1).

The American College of Radiology (ACR) mammography accreditation program changed the practice of mammography in the mid-1980s, with recommendations for minimum standards of practice and quality control (QC) that spurred improvements in technology and ensured quality of service. The federal Mammography Quality Standards Act (MQSA) was enacted in 1992. The law, and associated federal regulations (Title 21 of the Code of Federal Regulations, Part 900) issued by the US Food and Drug Administration (FDA), made many of the standards of the accreditation program mandatory. For digital mammography systems, many of the regulatory requirements entail following the manufacturer's recommended QC procedures.

Breast cancer screening programs depend on x-ray mammography because it is a low-cost, low–radiation dose procedure that has the sensitivity to detect early- stage breast cancer. Mammographic features characteristic of breast cancer are masses, particularly ones with irregular or "spiculated" margins; clusters of microcalcifications; and architectural distortions of breast structures. In screening mammography as practiced in the United States, two x-ray images of each breast, in the mediolateral oblique and craniocaudal views, are acquired. Whereas *screening mammography* attempts to identify breast cancer in the asymptomatic population, *diagnostic mammography* procedures are performed to assess palpable lesions or evaluate suspicious findings identified by screening mammography. The diagnostic mammographic examination may include additional x-ray projections, magnification views, spot compression views, ultrasound, magnetic resonance imaging (MRI), or mammoscintigraphy.

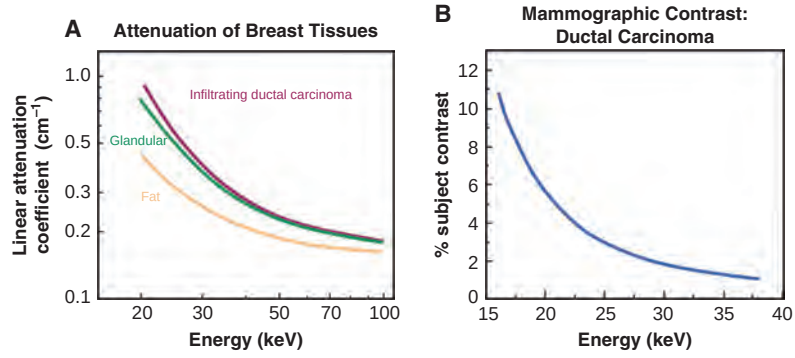
Ultrasound (Chapter 14) is often used to differentiate cysts (typically benign) from solid masses (often cancerous) and is also used when possible for biopsy needle guidance. MRI (Chapters 12 and 13) has excellent tissue contrast sensitivity and with contrast enhancement can differentiate benign from malignant tumors; it is used for diagnosis, staging, biopsy guidance, and, in some cases, screening. The clinical utility of mammoscintigraphy utilizing Tc-99m sestamibi is in the evaluation



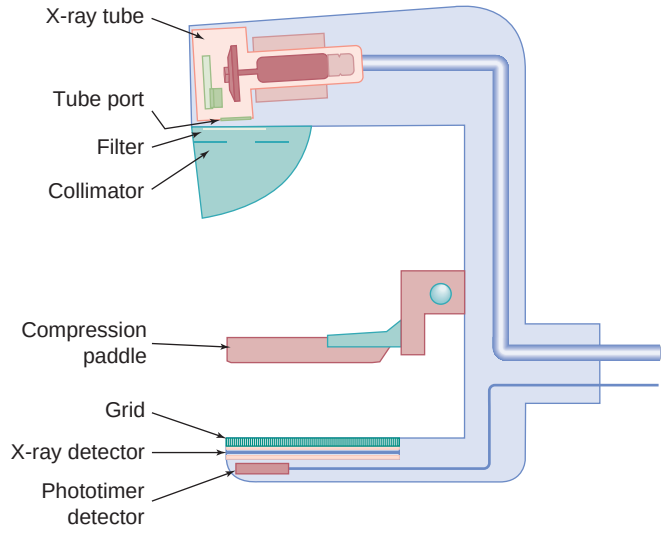
■ **FIGURE 8-1** Improvements in mammography. Images of a medial-lateral oblique view of the breast are shown. On the left is a state-of-the-art screen-film image (2006); in the middle is a corresponding digital mammogram (2008) demonstrating excellent exposure latitude and the skin line not seen in the film image; and on the right is one of approximately 50 digital tomosynthesis images (2010), which reduce superposition of anatomy in the projection image at about the same dose as a screen-film mammogram. Today, digital imaging equipment is used in the majority of mammography clinics.

of suspected breast cancer in patients for whom mammography is nondiagnostic, equivocal, or difficult to interpret (e.g., the presence of scar tissue, mammographically dense breast tissue, implants, or severe dysplastic disease). It is also used to assist in identifying multicentric and multifocal carcinomas in patients with tissue diagnosis of breast cancer.

The morphological differences between normal and cancerous tissues in the breast and the presence of microcalcifications require the use of x-ray equipment designed specifically to optimize breast cancer detection. As shown in Figure 8-2A, the attenuation differences between normal and cancerous tissue are extremely small. Subject contrast, shown in Figure 8-2B, is highest at low x-ray energies (10 to 15 keV) and reduced at higher energies (e.g., greater than 30 keV). Low x-ray



■ **FIGURE 8-2** **A.** Attenuation of breast tissues as a function of energy, showing fat, glandular, and ductal carcinoma linear attenuation coefficients. Comparison of the three tissues shows a very small difference between the glandular and the cancerous tissues. **B.** Calculated percentage contrast of the ductal carcinoma relative to the glandular tissue declines rapidly with energy; contrast is optimized by the use of a low-energy, nearly monoenergetic x-ray spectrum (Adapted from Yaffe MJ. Digital mammography. In: Haus AG, Yaffe MJ, eds. *Syllabus: a categorical course in physics: technical aspects of breast imaging*. Oak Brook, IL: RSNA, 1994:275–286.)



■ **FIGURE 8-3** A dedicated mammography system has many unique attributes. Major components of a typical system, excluding the generator and user console, are shown.

energies provide the best differential attenuation between the tissues; however, the high absorption results in higher radiation doses and long exposure times. Detection of small calcifications in the breast is also important because microcalcifications are in some cases early markers of breast cancer. Thus, mammography requires x-ray detectors with high spatial resolution that function best at higher doses. Enhancing contrast sensitivity, reducing dose, and providing the spatial resolution necessary to depict microcalcifications impose extreme requirements on mammographic equipment and detectors. Therefore, dedicated x-ray equipment, specialized x-ray tubes, breast compression devices, antiscatter grids, x-ray detectors (screen-film or digital), and phototimer detector systems are essential for mammography (Fig. 8-3). Strict QC procedures and cooperation among the technologist, radiologist, and medical physicist are necessary to ensure that high-quality mammograms are achieved at the lowest possible radiation doses. With these issues in mind, the technical and physical considerations of mammographic x-ray imaging are discussed below. Many of the basic concepts regarding image quality, x-ray production, and radiography are presented in Chapters 4, 6, and 7, respectively. The applications of these concepts to mammography, together with many new topics specific to mammography, are presented below.

8.1 X-ray Tube and Beam Filtration

A dedicated mammography x-ray tube has more similarities than differences when compared to a conventional x-ray tube. Therefore, this section highlights the differences between mammography and general radiography x-ray tubes.

Cathode and Filament Circuit

The mammography x-ray tube is configured with dual filaments in the focusing cup to produce 0.3- and 0.1-mm focal spot sizes, with the latter used for magnification studies to reduce geometric blurring, as discussed later in this chapter. An important

distinction between mammography and conventional x-ray tube operation is the low operating voltage, below 40 kV, which requires feedback circuits in the x-ray generator to adjust the filament current as a function of kV to deliver the desired tube current because of the nonlinear relationship between filament current and tube current as described in Chapter 6. In addition, the filament current is restricted to limit the tube current, typically to 100 mA for the large (0.3 mm) focal spot and 25 mA for the small (0.1 mm) focal spot so as to not overheat the Mo or Rh targets due to the small interaction areas. Higher filament currents and thus tube currents, up to and beyond 200 mA for the large focal spot and 50 mA for the small focal spot, are possible with tungsten anodes chiefly due to a higher melting point compared to Mo and Rh anodes.

Anode

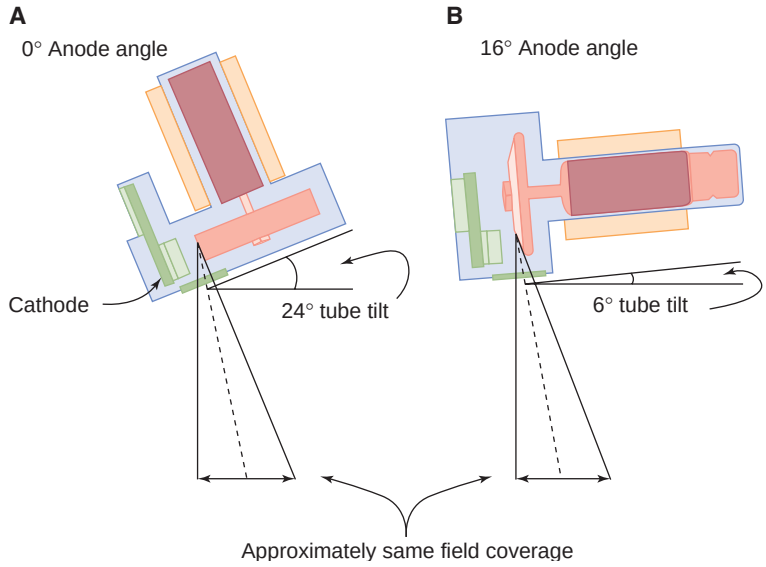
Molybdenum is the most common anode target material used in mammography x-ray tubes, but Rh and increasingly tungsten (W) are also used as targets. Characteristic x-ray production is the major reason for choosing Mo (K-shell x-ray energies of 17.5 and 19.6 keV) and Rh (20.2 and 22.7 keV) targets, as the numbers of x-rays in the optimal energy range for breast imaging are significantly increased by characteristic x-ray emission. With digital detectors, W is becoming the target of choice. Increased x-ray production efficiency, due to its higher atomic number, and improved heat loading, due to its higher melting point, are major factors in favor of W. Digital detectors have extended exposure latitude, and because post-acquisition image processing can enhance contrast, characteristic radiation from Mo or Rh is not as important in digital mammography as it is with screen-film detectors.

Mammography x-ray tubes have rotating anodes, with anode angles ranging from 16 to 0 degrees, depending on the manufacturer. The tubes are typically positioned at a source-to-image receptor distance (SID) of about 65 cm. In order to achieve adequate field coverage on the anterior side of the field, the x-ray tube must be physically tilted so that the *effective anode angle* (the actual anode angle plus the physical tube tilt) is at least 22 degrees for coverage of the 24 × 30-cm field area. A tube with a 0-degree anode angle requires a tube tilt of about 24 degrees to achieve an effective anode angle of 24 degrees (Fig. 8-4A). A 16-degree anode angle requires a tube tilt of 6 degrees for an effective angle of 22 degrees (Fig. 8-4B).

The intensity of the x-rays emitted from the focal spot varies within the beam, with the greatest intensity on the cathode side of the projected field and the lowest intensity on the anode side, a consequence of the *heel effect* (Chapter 6). Positioning the cathode over the chest wall of the patient and the anode over the anterior portion (nipple) achieves better uniformity of the *transmitted* x-rays through the breast (Fig. 8-5). Orientation of the tube in this way also decreases the equipment bulk near the patient's head. The anode is kept at ground potential (0 voltage), and the cathode is set to the highest negative voltage to reduce *off-focal radiation* (Chapter 6).

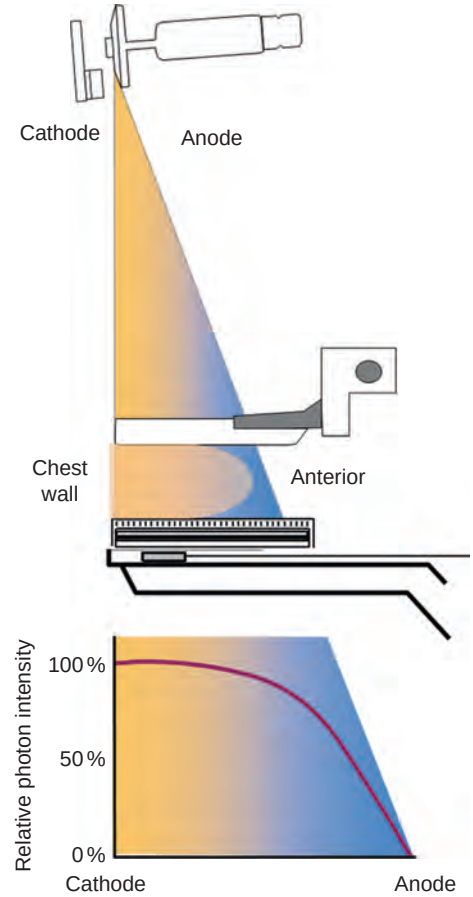
Focal Spot Considerations

Focal spot nominal sizes of 0.3 to 0.4 mm for contact mammography (breast compressed against the grid and image receptor) and 0.10 to 0.15 mm for magnification imaging (breast compressed against a magnification stand, which supports the breast at a distance from the image receptor to provide geometric image magnification)



■ **FIGURE 8-4** Design of a dedicated mammography system includes unique geometry and x-ray tube design. A "half-field" geometry is implemented with the x-ray beam central axis perpendicular to the plane of the image receptor at the center edge of the field at the chest wall, and centered horizontally. To provide full-area x-ray beam coverage over the large FOV (24 × 30 cm) at the 60- to 70-cm SID, a tube tilt of about 20 to 24 degrees is necessary. **A.** An anode angle of 0 degrees requires a tube tilt of 24 degrees. **B.** An anode angle of 16 degrees requires a tube tilt of about 6 degrees.

■ **FIGURE 8-5** Orientation of the cathode-anode direction of the x-ray tube is from the chest wall side of the breast (over the cathode) to the anterior side of the breast (over the anode). Variation of x-ray beam intensity is caused by the heel effect, a result of anode self-filtration of x-rays directed toward the anode side of the field, with a significant reduction in x-ray intensity. The thickest part of the breast (at the chest wall) is positioned below the cathode, which helps equalize x-ray intensities reaching the image receptor that are transmitted through the breast in the cathode-anode direction.



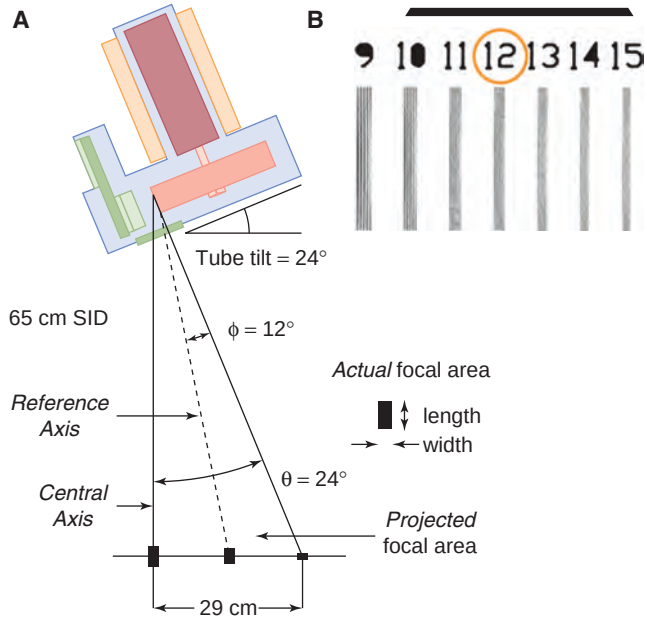
reduce geometric blurring so that microcalcifications can be resolved. A consequence of using small focal spots is reduced maximal tube current (e.g., for a Mo target, 100 mA for the large focal spot and 25 mA for the small focal spot) and correspondingly longer exposure times.

In order to avoid exposure of the patients' torsos and to maximize the amount of breast tissue near the chest wall that is imaged, all dedicated mammography systems utilize a "half-field" x-ray beam geometry, which is achieved by fixed collimation at the x-ray tube head. As a result, the central axis of the x-ray beam is directed at the chest wall edge of the receptor and perpendicular to the plane of the image receptor. Furthermore, by convention, the nominal focal spot size is measured at the *reference axis*, which bisects the x-ray field along the chest wall—anterior direction of the x-ray field (Fig. 8-6A).

The sum of the target angle and tube tilt is equal to θ, and ϕ represents the reference angle. At the reference axis, the focal spot length, a_{ref}, is foreshortened compared to the focal spot length at the chest wall, $a_{\text{chest wall}}$, given by the relationship

$$a_{ref} = a_{\text{chest wall}} \left(1 - \frac{\tan(\theta - \phi)}{\tan \theta} \right) \qquad [8\text{-}1]$$

Nominal focal spot size and tolerance limits for the width and length are listed in Table 8-1. As a consequence of the effective focal spot size variation in the field (the line focus principle), sharper image detail is rendered on the anode (anterior) side of the field toward the nipple, which is more evident with magnification examinations, as described in Section 8.3.



■ **FIGURE 8-6** **A.** The projected focal spot size varies along the cathode-anode axis, and the nominal focal spot size is specified at a *reference* axis at an angle ϕ from a line perpendicular to the cathode-anode axis (*dashed line*, ϕ = 12 degrees). The actual focal area is the electron distribution area on the anode (width × length), and the projected length is foreshortened according to the line focus principle. **B.** A resolution bar pattern with object magnification measures overall system resolution including that of the focal spot. Bars of 12 line pairs per mm are resolved for this particular system (magnification = 1.5×, 0.1-mm focal spot).

TABLE 8-1 NOMINAL FOCAL SPOT SIZE AND MEASURED TOLERANCE LIMITS OF MAMMOGRAPHY X-RAY TUBES SPECIFIED BY NEMA AND MQSA REGULATIONS

NOMINAL FOCAL SPOT SIZE (mm)	WIDTH (mm)	LENGTH (mm)
0.10	0.15	0.15
0.15	0.23	0.23
0.20	0.30	0.30
0.30	0.45	0.65
0.40	0.60	0.85
0.60	0.90	1.30

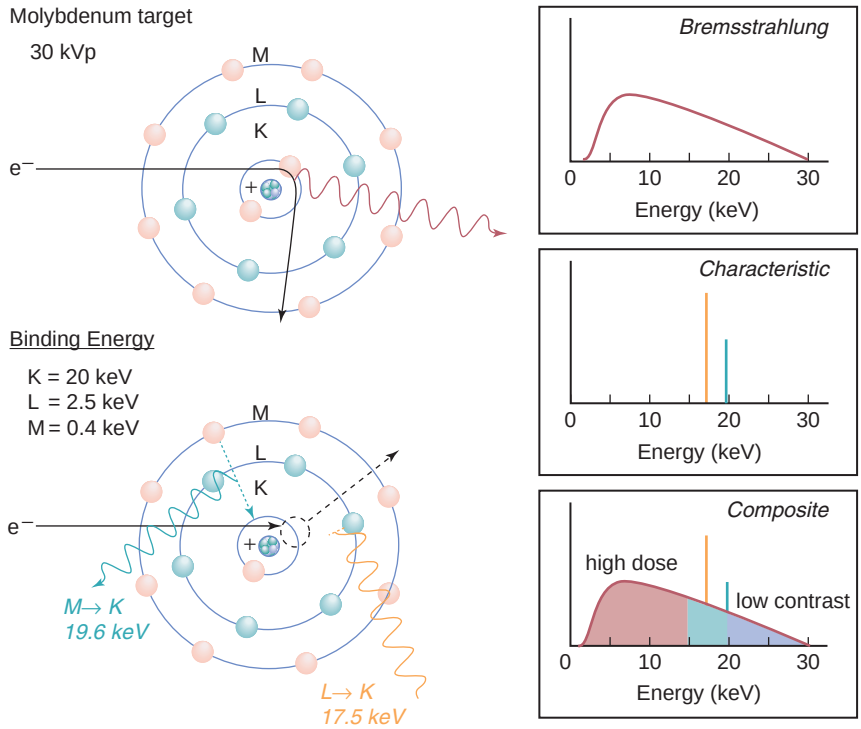
National Electrical Manufacturers Association publication XR-5, and MQSA regulations 21 CFR 900.

Focal spot resolvability with or without magnification is measured by imaging a high-resolution bar pattern with up to 20 line pairs/mm. For conventional breast imaging, the bar pattern is placed 4.5 cm above the breast support surface near the chest wall; for magnification imaging, it is placed 4.5 cm above the magnification platform. Orientation of the bar pattern parallel and perpendicular to the cathode-anode direction yields measurements of overall *effective* resolution that reflect the effects of the focal spot dimensions, the acquisition geometry, and the detector sampling characteristics. A resolution bar pattern is shown in Figure 8-6B for a magnification study using the 0.1-mm focal spot. The resolving capability of the imaging system is limited by the component that causes the most blurring. In magnification mammography, this is generally the focal spot, whereas in contact mammography, it may be the detector element size, and at other times patient motion.

Tube Port, Tube Filtration, and Beam Quality

The tube port and added tube filters play an important role in shaping the mammography x-ray energy spectrum. The tube port window is made of beryllium. The low atomic number (Z=4) of beryllium and the small thickness of the window (0.5 to 1 mm) allow the transmission of all but the lowest energy (less than 5 keV) bremsstrahlung x-rays. In addition, Mo and Rh targets produce beneficial K-characteristic x-ray peaks at 17.5 and 19.6 keV (Mo) and 20.2 and 22.7 keV (Rh) (see Chapter 6, Table 6-2), whereas tungsten targets produce a large fraction of unwanted L-characteristic x-rays at 8 to 10 keV. Figure 8-7 shows a bremsstrahlung, characteristic, and composite x-ray spectrum from an x-ray tube with a Mo target and Be window operated at 30 kV.

Added x-ray tube filtration improves the energy distribution of the mammography output spectrum by selectively removing the lowest and highest energy x-rays from the x-ray beam, while largely transmitting desired x-ray energies. This is accomplished by using elements with *K-absorption edge* (Chapter 3) energies between 20 and 27 keV. Elements that have these K-shell binding energies include Mo, Rh, and Ag, and each can be shaped into thin, uniform sheets to be used as added x-ray tube filters. At the lowest x-ray energies, the attenuation of added filtration is very high. The attenuation decreases as the x-ray energy increases up to the K-edge of the filter element. For x-ray energies just above this level, photoelectric absorption interactions dramatically increase attenuation as a step or "edge" function (Fig 8-8A). At higher x-ray energies, the attenuation decreases. The result is the selective transmission of x-rays in a narrow band of energies from about 15 keV up to the K-absorption edge of the filter.

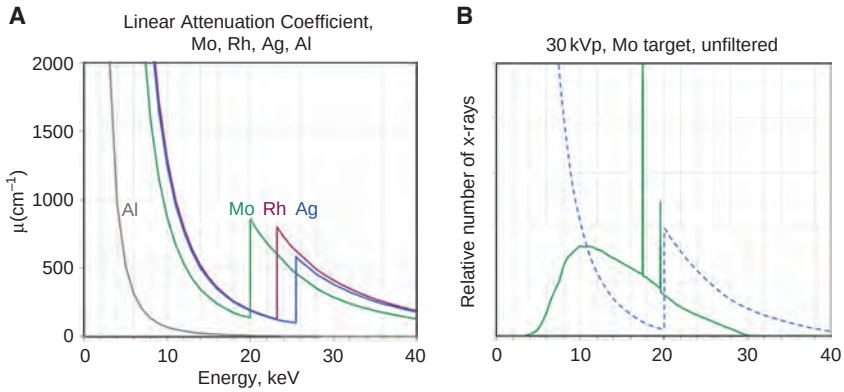


■ **FIGURE 8-7** The x-ray spectrum of a mammography x-ray tube is composed of bremsstrahlung (with a continuous photon energy spectrum) and characteristic (discrete energies) radiation (see Chapter 6 for more details). A Mo anode tube operated at 30 kV creates the continuous spectrum as well as characteristic radiation with photon energies of 17.5 and 19.6 keV. On the lower right, the "unfiltered" composite spectrum transmitted through 1 mm of Be (tube port material) has a large fraction of low-energy x-rays that will deposit high dose without contributing to the image, and a substantial fraction of high-energy x-rays that will reduce subject contrast of the breast tissues. The ideal spectral energy range is from about 15 to 25 keV, depending on breast tissue composition and thickness.

In Figure 8-8B, the unfiltered Mo target spectrum and a superimposed attenuation curve for a Mo filter are shown. Importantly, the characteristic x-ray energies produced by the Mo target occur at the lowest attenuation of the filter in this energy range.

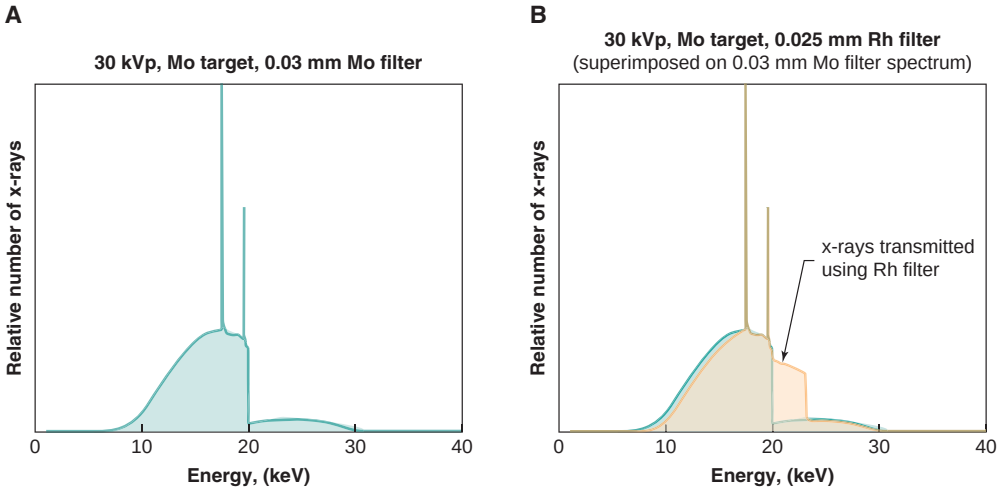
With a Mo target, a 0.030-mm-thick Mo filter or a 0.025-mm Rh filter is typically used, and for a Rh target, a 0.025-mm Rh filter is used. A variety of filters are used with W targets, including Rh (0.05 mm), Ag (0.05 mm), and Al (0.7 mm).

The spectral output of a Mo target and 0.030-mm-thick Mo filter is shown in Figure 8-9A, illustrating the selective transmission of Mo characteristic radiation and significant attenuation of the lowest and highest x-rays in the transmitted spectrum. Tuning the spectrum to achieve optimal effective x-ray energy for breast imaging is accomplished by selecting the anode material, added filtration material, and kV. Screen-film detectors most often use a Mo target and 0.03-mm Mo filtration with a kV of 24 to 25 kV for thin, fatty breasts and up to 30 kV for thick, glandular breasts. For thicker and denser breasts, a Mo target and Rh filter are selected with higher voltage, from 28 to 32 kV, to achieve a higher effective energy and more penetrating beam (Fig. 8-9B, right graph). Some systems have Rh targets, which produce Rh characteristic x-ray energies of 20.2 and 22.7 keV, achieving an even higher effective energy x-ray beam for a set kV (Fig. 8-10A). A Mo filter should *never* be used with a Rh target, because Rh characteristic x-rays are attenuated significantly as their energies are above the Mo K-absorption edge (Fig. 8-10B).

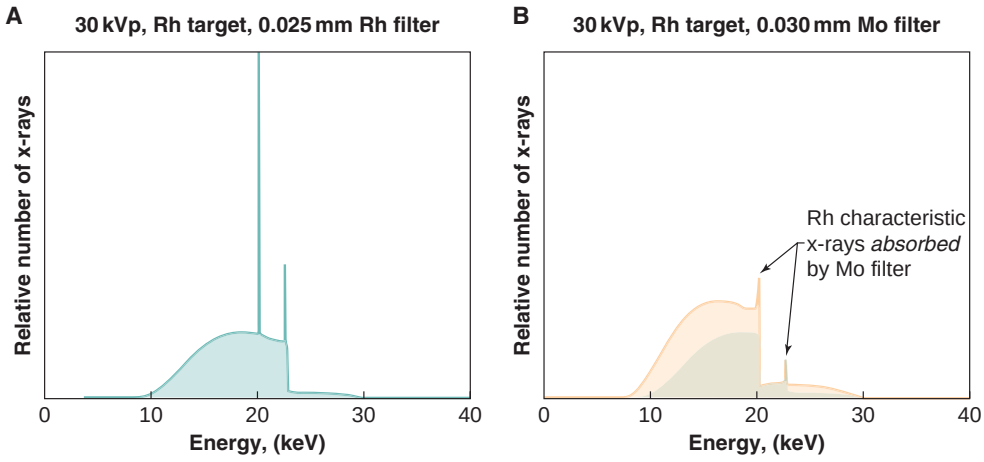


■ **FIGURE 8-8** **A**. The linear attenuation coefficients of Al, Mo, Rh, and Ag are plotted as a function of energy. A low-attenuation "window" exists below the *K*-absorption edge energy for the higher Z elements. **B**. An unfiltered Mo target spectrum generated at 30 kV shows a large fraction of low- and high-energy photons. Superimposed at the same energy scale is a *dashed line* illustrating a Mo filter attenuation as a function of energy.

W targets are now used for many digital mammography systems because of their higher bremsstrahlung production efficiency and higher tube loadings than Mo and Rh targets. *K*-edge filters can optimize the output energy spectrum for breast imaging. However, an unfiltered W spectrum contains a huge fraction of unwanted L x-rays in the 8- to 12-keV range (Fig. 8-11A). Therefore, minimum filter thicknesses of 0.05 mm for Rh and Ag are needed to attenuate the L-x-rays to negligible levels, as shown in Figure 8-11B. In some applications, an Al filter is used, but because of its low Z and lack of a useful K-absorption edge, a thickness of 0.7 mm is necessary to attenuate the L x-rays. In digital tomosynthesis, in which contrast is largely provided by the tomographic image reconstruction process (Section 8.5), breast dose can be reduced by the use of Al filtration, which yields a larger fraction of high-energy x-rays.



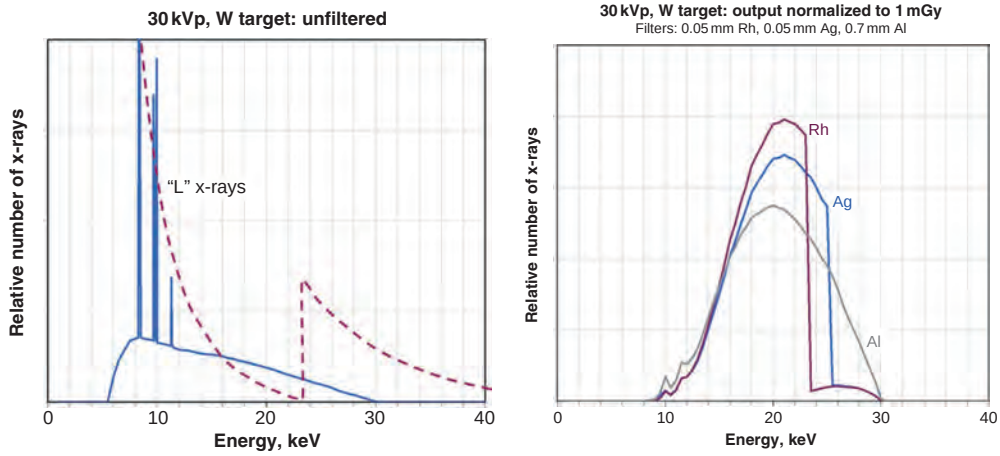
■ **FIGURE 8-9** **A**. A filtered output spectrum is shown for a Mo target and 0.030-mm Mo filter for a 30-kV tube voltage, with prominent characteristic peaks at 17.5 and 19.6 keV. This is a common spectrum for less dense and thinner breast tissues, particularly with lower operating tube voltages (25 to 28 kV). **B**. A filtered output spectrum is shown for a Mo target and 0.025-mm Rh filter, superimposed on the Mo/Mo spectrum in (A), indicating the transmission of higher energy bremsstrahlung x-rays up to the K absorption edge of Rh at 23.2 keV. The spectrum, including the Mo characteristic x-rays, is more energetic and penetrating, and is preferable for imaging thicker and denser breast tissues.



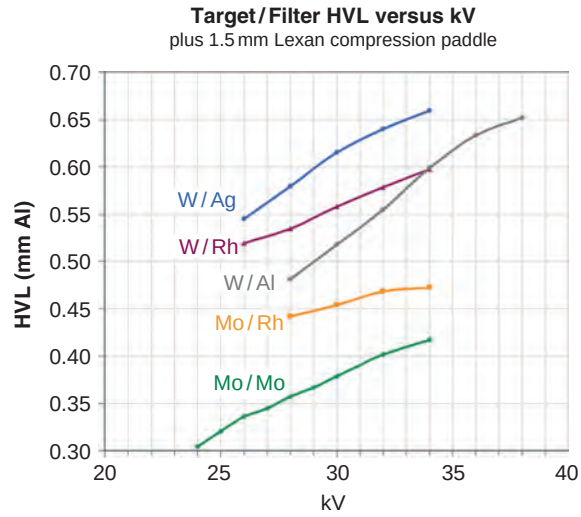
■ **FIGURE 8-10** **A.** A filtered output spectrum is shown for a Rh target and 0.025-mm Rh filter for a 30-kV tube voltage, with prominent Rh characteristic x-rays at 20.2 and 22.7 keV. This spectrum provides a higher effective energy than a Mo target—Rh filter x-ray beam at the same kV, due to the higher energy characteristic x-rays. **B.** A combination of a Rh target and Mo filter *is inappropriate*, as the Rh characteristic x-rays are strongly attenuated as shown.

Half-Value Layer

The half-value layer (HVL) of a mammography x-ray beam ranges from 0.3 to 0.7-mm Al for the kV range and combinations of target material, filter material, and filter thickness used in mammography (Fig. 8-12). The HVL depends on the target material (Mo, Rh, W), kV, filter material, and filter thickness. Measurement of the HVL is usually performed with the compression paddle in the beam, using 99.9% pure Al sheets of 0.1-mm thickness. A Mo target, 0.03-mm Mo filter, 28 kV, and a 1.5 mm Lexan compression paddle produce a HVL of about 0.35-mm Al, whereas a W target, 0.05-mm Rh filter, and 30 kV produce a HVL of about 0.55-mm Al. HVLs vary



■ **FIGURE 8-11** **A.** An unfiltered W target spectrum generated at 30 kV shows a large fraction of lower energy L-Characteristic x-rays at 8 to 11.5 keV. Superimposed at the same energy scale is a *dashed line* illustrating the high- and low-attenuation characteristics of a Rh filter. **B.** Filtered W-target x-ray spectra by Rh (0.05 mm), Ag (0.05 mm), and Al (0.7 mm) are normalized to the same exposure output.



■ **FIGURE 8-12** The HVL (including the compression paddle attenuation) versus kV is plotted for Mo targets with Mo and Rh filters, and for W targets with Rh, Ag, and Al filters. HVL measurements are representative of an average of a number of evaluations on various mammography systems at the UC Davis Health System and are generally within +/− 0.05 mm of the values shown.

from machine to machine because of slight variation in actual filter thicknesses and kV. The HVL of breast tissue is highly dependent on tissue composition (glandular, fibrous, or fatty) and the HVL of the incident x-ray beam. Usually, the HVL for breast tissues is from 1 to 3 cm.

Minimum HVL limits, listed in Table 8-2, are prescribed by MQSA regulations to ensure that a beam of adequate effective energy is achieved. Maximum HVL limits are recommended by the ACR accreditation guidelines, as listed in Table 8-3. An x-ray beam that is "harder" than optimal indicates too much filtration or a pitted anode or aged tube and can result in reduced output and poor image quality. Estimates of the radiation dose to the breast require accurate assessment of the HVL (see Section 8.6).

Tube Output and Tube Output Rate

Tube output is a measure of the intensity of the x-ray beam, typically normalized to mAs or to 100 mAs, at a specified distance from the source (focal spot). Common units of tube output are mGy (air kerma)/100 mAs and mR (exposure)/mAs. The kV, target, filter material and thickness, distance from the source, and focal spot size must be specified. Figure 8-13 shows the output at a 50-cm distance from Mo and W

TABLE 8-2 REQUIREMENTS FOR MINIMUM HVL IN THE MQSA REGULATIONS (21 CFR PART 1020.30); ACR: WITH COMPRESSION PADDLE

TUBE VOLTAGE (kV)	MQSA: kV/100	ACR: kV/100 + 0.03
24	0.24	0.27
26	0.26	0.29
28	0.28	0.31
30	0.30	0.33
32	0.32	0.35

ACR, American College of Radiology; CFR, Code of Federal Regulations; MQSA, Mammography Quality Standards Act.

TABLE 8-3 ACR SPECIFICATIONS FOR MAXIMUM HVL: MAXIMUM HVL (mm Al) = kV/100 + *C* FOR TARGET-FILTER COMBINATIONS

TUBE VOLTAGE (kV)	Mo/Mo *C* = 0.12	Mo/Rh *C* = 0.19	Rh/Rh *C* = 0.22	W/Rh *C* = 0.30
24	0.36	0.43	0.46	0.54
26	0.38	0.45	0.48	0.56
28	0.40	0.47	0.50	0.58
30	0.42	0.49	0.52	0.60
32	0.44	0.51	0.54	0.62

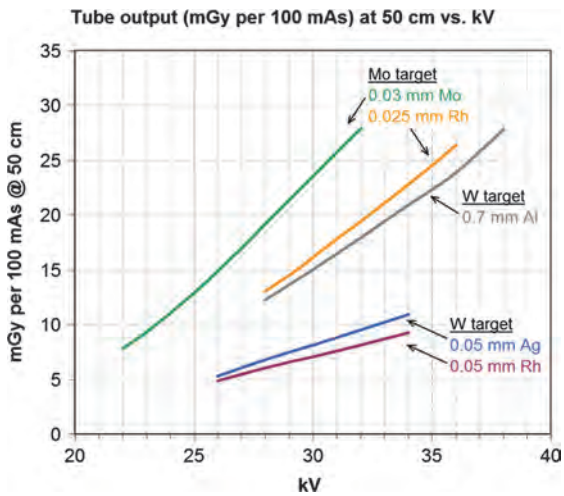
ACR Accreditation Program requirements as specified in the Medical Physics testing section, http://www.acr.org

target x-ray tubes with a variety of tube filter materials and thicknesses. Even though W targets are more efficient at producing x-rays, the thicker filters needed to attenuate the L-characteristic x-rays result in lower tube output per mAs compared to the Mo target. However, W spectra have higher HVLs and greater beam penetrability, allow higher tube current, and result in comparable exposure times to a Mo target and filter for a similar breast thickness.

X-ray tube output values are useful for calculating the free-in-air incident air kerma (or exposure) to the breast for a mammography system's target and filter combination, kV, mAs, and source-to-breast surface distance. The source-to-breast surface distance is determined from the known SID, breast platform to detector distance, and compressed breast thickness. For instance, assume that a mammography system with a Mo target and Rh filter uses 30 kV and 160 mAs for a SID of 65 cm, compressed breast thickness of 6 cm, and a breast platform to detector distance of 2 cm. The entrant breast surface to the source is closer by 8 cm, and is therefore 57 cm from the source. From Figure 8-13, the tube output is 16 mGy/100 mAs for 30 kV at a distance of 50 cm. Calculation of incident air kerma considers tube output at a specific kV, the mAs used, and inverse square law correction from 50 to 57 cm:

$$16 \text{ mGy}/100 \text{ mAs} \times 160 \text{ mAs} \times [50.0/57.0]^2 = 19.7 \text{ mGy}$$

Calculation of the average glandular dose to the breast is determined from the measured incident air kerma value and other parameters as discussed in Section 8.6.



■ **FIGURE 8-13** Tube output (mGy/100 mAs at 50-cm distance from the source with compression paddle in the beam) for two clinical mammography units is measured at 2-kV intervals, and the data are smoothed. A Mo target with 0.030-mm Mo and 0.025-mm Rh filter, and a W target with 0.05-mm Rh, 0.05-mm Ag, and 0.7-mm Al filters are plotted.

Tube output rate is the air kerma rate at a specified distance from the x-ray focal spot and is a function of the tube current achievable for an extended exposure time (typically ~300 mAs for an exposure time greater than 3 s). To ensure the ability to deliver a sufficient x-ray beam fluence rate to keep exposure times reasonable, MQSA regulations require that systems be capable of producing an air kerma rate of at least 7.0 mGy/s, when operating at 28 kV in the standard (Mo/Mo) mammography mode, at any SID for which the system is designed to operate for screen-film detectors. Digital systems have requirements specified by the manufacturers' QC manuals.

X-ray Tube Alignment and Beam Collimation

Alignment of the x-ray tube and collimator are crucial to ensure that the x-ray beam central axis is perpendicular to the plane of the image receptor and intercepts the center of the chest wall edge of the image receptor. This will protect the patient from unnecessary radiation exposure to the lungs and torso and include as much breast tissue as possible in the image. Tube alignment with respect to the image receptor must be verified during acceptance testing of the system and after x-ray tube replacements.

Collimation of the x-ray beam is achieved by the use of fixed-size metal apertures or adjustable shutters. For most screen-film examinations, the field size is automatically set to match the film cassette dimensions (e.g., 18 × 24 cm or 24 × 30 cm). For a large area digital detector (24 × 30 cm), when operating with the smaller field area (18 × 24 cm), one of three active acquisition areas is used: center, left shift, and right shift, which requires the collimator assembly to restrict the x-ray field to the corresponding active detector area. Field shifts are used for optimizing the oblique projections for subjects with smaller breasts to accommodate positioning of the arm and shoulder at the top edge of the receptor. Collimation to the full active detector area is the standard of practice (see Section 8.4: "Film-Viewing Conditions"). There is no disadvantage to full-field collimation compared to collimation to the breast only, as the tissues are fully in the beam in either case. However, for magnification and spot compression studies, manually adjusted shutters allow the x-ray field to be more closely matched to the volume being imaged.

The projected x-ray field must extend to the chest wall edge of the image receptor without cutoff, but not beyond the receptor by more than 2% of the SID. If the image shows evidence of collimation of the x-ray field on the chest wall side or if the chest wall edge of the compression paddle is visible in the image, service must be requested. A collimator light and mirror assembly visibly display the x-ray beam area. Between the tube port and the collimator is a low-attenuation mirror that directs light from the collimator lamp through the collimator opening to emulate the x-ray field area. Similar to conventional x-ray tube collimator assemblies, the light field must be congruent with the actual x-ray field to within 2% overall, which is achieved by adjustment of the light and mirror positions. MQSA regulations require repairs to be made within 30 days if the light/x-ray field congruence, x-ray field-image receptor alignment, or compression paddle alignment is out of tolerance as described above.

8.2 X-ray Generator and Phototimer System

X-ray Generator

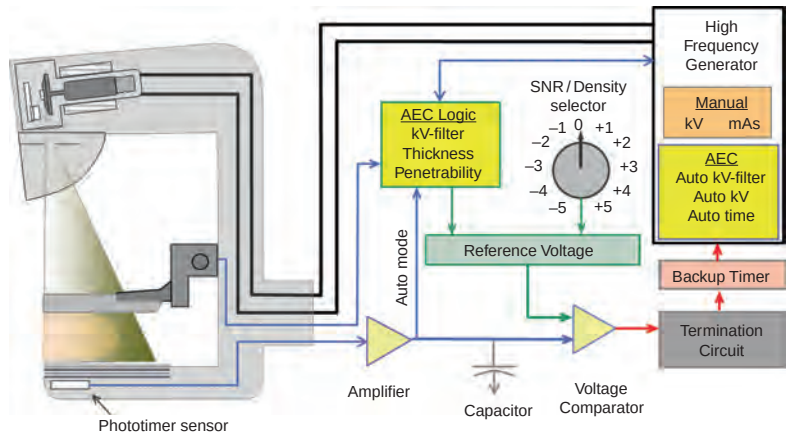
A dedicated mammography x-ray generator is similar to a conventional x-ray generator in design and function. Differences include the lower voltages supplied to the x-ray tube, space charge compensation, and automatic exposure control (AEC)

circuitry. In most cases, AEC is employed for mammography screening and diagnostic examinations, although there are instances when the technologist will use manual controls to set the tube current - exposure duration product (mAs). Like most contemporary x-ray imaging systems, high-frequency generators are used for mammography due to low voltage ripple, fast response, easy calibration, long-term stability, and compact size. There are circuits that prohibit x-ray exposures if the setup is incomplete (e.g., insufficient compression, cassette not in the tunnel, or digital detector not ready to capture an image).

Automatic Exposure Control

The AEC employs a radiation sensor or sensors, a charge amplifier, and a voltage comparator to control the exposure (Fig. 8-14). For cassette-based image receptors (screen-film *and* computed radiography [CR]), the phototimer sensor is located *underneath* the cassette, and consists of a single ionization chamber or an array of three or more semiconductor diodes. X-rays transmitted through the breast, antiscatter grid (if present), and the image receptor generate a signal in the detector. The signal is accumulated (integrated) and, when the accumulated signal reaches a preset value, the exposure is terminated. The preset value corresponds to a specified signal-to-noise ratio (SNR) in a digital mammography unit or an acceptable optical density (OD) if a film-screen system is used. For an active matrix flat-panel imager, the detector array itself can be used to measure the x-rays transmitted through the breast and the antiscatter grid.

Phototimer algorithms use several inputs to determine the radiographic technique, including compressed breast thickness, phototimer adjustment settings on the generator console, the kV, the tube anode selection (if available), and the tube filter to achieve the desired film OD or digital SNR in the acquired image. The operator typically has two or three options for AEC: (a) a fully automatic AEC mode that sets the optimal kV and filtration (and target material on some systems) from a short test exposure of approximately 100 ms to determine the penetrability of the breast; (b) automatic kV with a short test exposure, with user-selected target and filter values; and (c) automatic time of exposure using manually set target, filter, and



■ **FIGURE 8-14** The AEC (phototimer) circuits use algorithms with input from the breast thickness (from the compression paddle), breast density (via a test exposure), density setting (operator density selector), and exposure duration (reciprocity law failure issues) to determine overall exposure time. In fully automatic modes, the kV, beam filter, and target are also determined, usually from a short (100 ms) exposure to determine attenuation characteristics of the breast.

kV values. For most patient imaging, the fully automatic mode is used, but, for QC procedures, the automatic exposure time adjustment mode with other parameters manually selected by the user is often employed.

At the generator control panel, an exposure adjustment selector modifies the AEC response to permit adjustments for unusual imaging circumstances such as imaging breasts with implants, magnification, or for radiologist preference in film OD or digital image SNR. Usually, eleven steps (−5 to 0 [neutral] to +5) are available to increase or decrease exposure by adjusting the AEC comparator switch. Each step provides a difference of 10% to 15% positive (greater exposure) or negative (lesser exposure) per step from the baseline (0) setting.

An x-ray exposure that is not stopped by the AEC circuit and exceeds a preset time (e.g., greater than 5 s) is terminated by a *backup timer*. This can occur if there is a malfunction of the AEC system or if the kV is set too low with insufficient x-ray transmission. In the latter situation, the operator must select a *higher* kV to achieve greater beam penetrability and shorter exposure time to correct the problem.

Inaccurate phototimer response, resulting in an under- or overexposed film or digital image, can be caused by breast tissue composition (adipose versus glandular) heterogeneity, compressed thickness beyond the calibration range (too thin or too thick), a defective cassette, a faulty phototimer detector, or an inappropriate kV setting. Screen-film response to very long exposure times, chiefly in magnification studies caused by the low mA capacity of the small focal spot, results in *reciprocity law failure* (Chapter 7), whereby the resultant OD is less than would be predicted by the amount of radiation delivered to the screen-film cassette; consequently, insufficient film OD results, and extension of the exposure time is added to compensate. For extremely thin or fatty breasts, the phototimer circuit and x-ray generator can be too slow in terminating the exposure, causing film overexposure. In situations where the exposure is terminated too quickly, grid-line artifacts will commonly appear in the image due to lack of complete grid motion during the short exposure.

The position of the phototimer detector (e.g., under dense or fatty breast tissue) can have a significant effect on film density but has much less of an impact on digital images because of postprocessing capabilities, even though the SNR will vary. Previous mammograms can aid the technologist in selecting the proper position for the phototimer detector to achieve optimal film density or optimal SNR for glandular areas of the breast. Most systems allow phototimer positioning in the chest wall to anterior direction, while some newer systems also allow side-to-side positioning to provide flexibility for unusual circumstances.

Technique Chart

Technique charts are useful guides to help determine the appropriate kV and target-filter combinations for specific imaging tasks, based on breast thickness and breast composition (fatty versus glandular tissue fractions). Most mammographic techniques use phototiming, and the proper choice of kV is essential for a reasonable exposure time, defined as a range from approximately 0.5 to 2.0 s to achieve an OD of 1.5 to 2.0 for film, or the desired SNR for digital images. Too short an exposure can cause visible grid lines to appear on the image, whereas too long an exposure can result in breast motion, either of which degrades the quality of the image. Table 8-4 lists kV recommendations for screen-film detectors to meet the desired exposure time range for breast thickness and breast composition using a Mo target and 0.030-mm Mo filter, determined from computer simulations and experimental

TABLE 8-4 RECOMMENDED kV, AS A FUNCTION OF BREAST COMPOSITION AND THICKNESS

	BREAST THICKNESS (cm)						
BREAST COMPOSITION	2	3	4	5	6	7	8
Fatty	24	24	24	24	25	27	30
50/50	24	24	24	25	28	30	32
Glandular	24	24	26	28	31	33	35

The goal is an exposure time between 0.5 and 2.0 s. This chart is for a Mo target and 0.030-mm Mo filter using a 100-mA tube current and screen-film detector. These kVs may not be appropriate for other target/filter combinations or for digital detectors. The kV values were determined by computer simulations and clinical measurements.

measurements. Techniques for digital detector systems are more flexible, and because of postprocessing and contrast enhancement capabilities, will often use a wider range of kV settings. A technique chart example for an amorphous selenium (Se) flat panel detector system using a Mo target, a W target, or a W target (without a grid for a digital tomosynthesis acquisition) and corresponding tube filtration is listed in Table 8-5.

8.3 Compression, Scattered Radiation, and Magnification

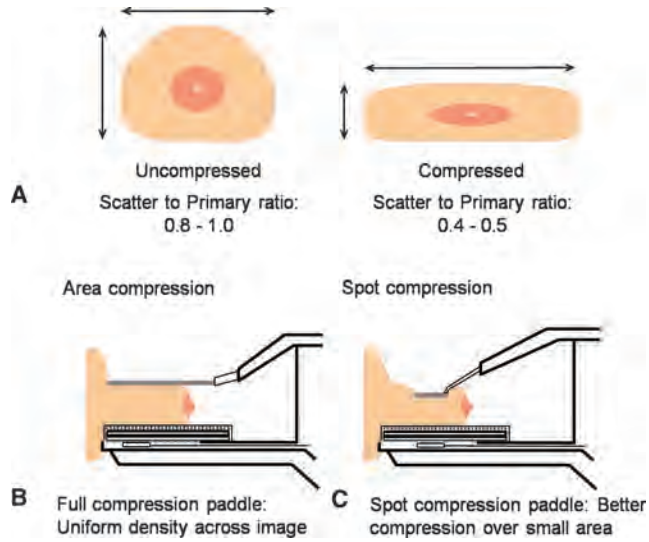
Compression

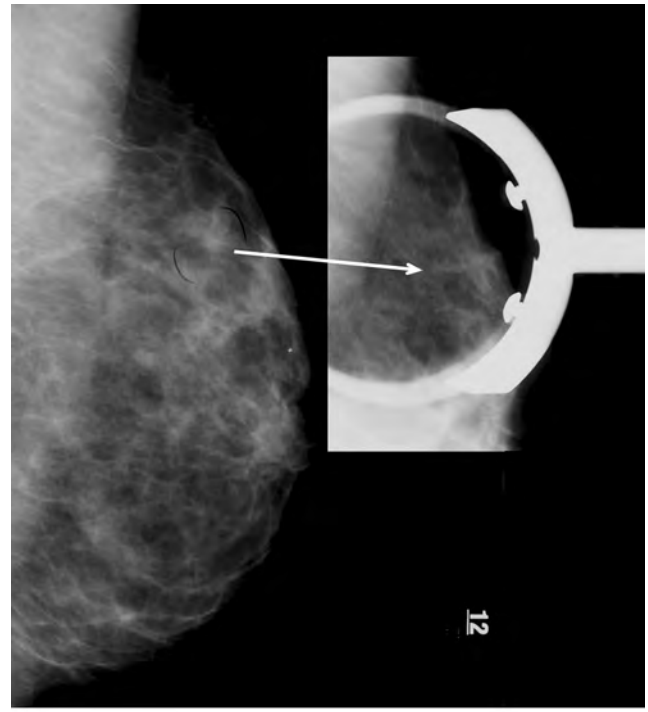
Breast compression is an important part of the mammography examination, whether using screen-film or digital detectors. Firm compression reduces overlapping anatomy, decreases tissue thickness, and reduces inadvertent motion of the breast (Fig. 8-15A). It results in fewer scattered x-rays, less geometric blurring of anatomic structures, and lower radiation dose to the breast tissues. Achieving a uniform breast thickness lessens exposure dynamic range and allows the use of higher contrast film, or allows more flexibility in image processing enhancement of the digital image.

Compression is achieved with a compression paddle, a Lexan plate attached to a mechanical assembly (Fig. 8-15B). The full area compression paddle matches the size of the image receptor (18 × 24 cm or 24 × 30 cm) and is flat and parallel to the breast support table. An alternative to the flat compression paddle is the "flex" paddle

TABLE 8-5 TECHNIQUE CHART FOR A DIGITAL MAMMOGRAPHY SYSTEM WITH SELENIUM DETECTOR FOR 50% GLANDULAR 50% ADIPOSE COMPOSITION

	DIGITAL, Mo TARGET, CELLULAR GRID			DIGITAL, W TARGET, CELLULAR GRID			DIGITAL TOMOSYNTHESIS, W TARGET, NO GRID		
THICKNESS (cm)	FILTER	kV	mAs	FILTER	kV	mAs	FILTER	kV	mAs
<3	Mo 30 μm	24	35	Rh 50 μm	25	40	Al 700 μm	26	35
3–5	Mo 30 μm	27	75	Rh 50 μm	28	80	Al 700 μm	29	50
5–7	Mo 30 μm	31	100	Rh 50 μm	31	150	Al 700 μm	33	65
>7	Rh 25 μm	33	150	Ag 50 μm	32	200	Al 700 μm	38	85





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■ **FIGURE 8-15** **A**. Compression is essential for mammographic studies to reduce breast thickness (less scatter, reduced radiation dose, and shorter exposure time) and to spread out superimposed anatomy. **B**. Suspicious areas often require "spot" compression to eliminate superimposed anatomy by further spreading the breast tissues over a localized area. **C**. Example of image with suspicious finding (**left**). Corresponding spot compression image shows no visible mass or architectural distortion (**right**).

that is spring loaded on the anterior side to tilt and accommodate variations in breast thickness from the chest wall to the nipple, providing more uniform compression of the breast. A compression paddle has a right-angle edge at the chest wall to produce a flat, uniform breast thickness when an adequate force of 111 to 200 newtons (25 to 44 lb) is applied. A smaller "spot" compression paddle (~5-cm diameter) reduces the breast thickness further over a specific region and redistributes the tissue for improved contrast and reduced anatomic overlap (Fig. 8-15C,D).

Typically, a hands-free (e.g., foot-pedal operated), motor-driven compression paddle is used, which is operable from both sides of the patient. In addition, a mechanical adjustment knob near the paddle holder allows fine manual adjustments of compression. While firm compression is not comfortable for the patient, it is often necessary for a clinically acceptable image.

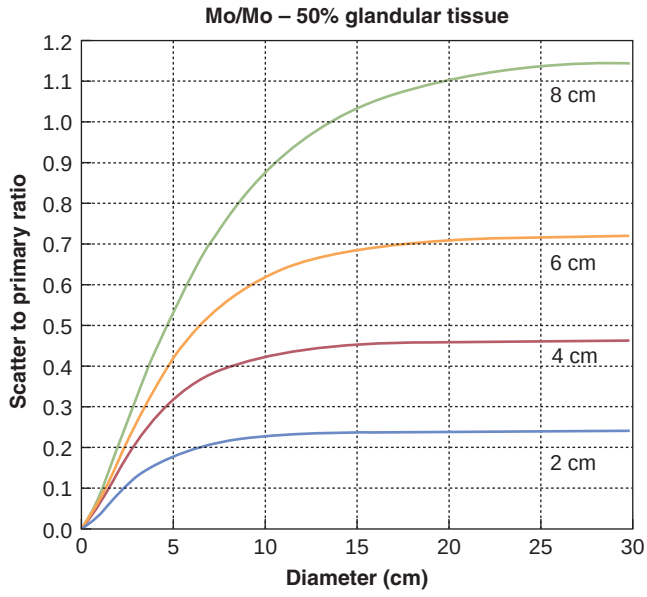
Scattered Radiation and Antiscatter Grids

X-rays transmitted through the breast contain primary and scattered radiation. Primary radiation carries information regarding the attenuation characteristics of the breast and delivers the maximum possible subject contrast to the detector. Scattered radiation is an additive, gradually varying radiation distribution that degrades subject contrast and adds random noise. If the maximum subject contrast without scatter is C_0 and the maximum contrast with scatter is C_S, then the contrast degradation factor (CDF) is approximated as

$$CDF = \frac{C_s}{C_0} = \frac{1}{1 + SPR} \qquad [8\text{-}2]$$

where SPR is the scatter-to-primary ratio (Chapter 7). X-ray scatter increases with increasing breast thickness and breast area, with typical SPR values shown in Figure 8-16. A breast of 6-cm compressed thickness will have an SPR of approximately 0.6 and a calculated scatter degradation factor of (1/(1+0.6)) = 0.625 = 62.5%. Without some form of scatter rejection, therefore, only a fraction of the inherent subject contrast can be detected.

In digital mammography, unlike screen-film mammography, the main adverse effect of scattered x-rays is not a reduction of contrast. Contrast can be improved by any desired amount by post-acquisition processing, such as windowing. The main adverse effect of scatter in digital mammography is that scattered photons add random noise, degrading the signal to noise ratio.

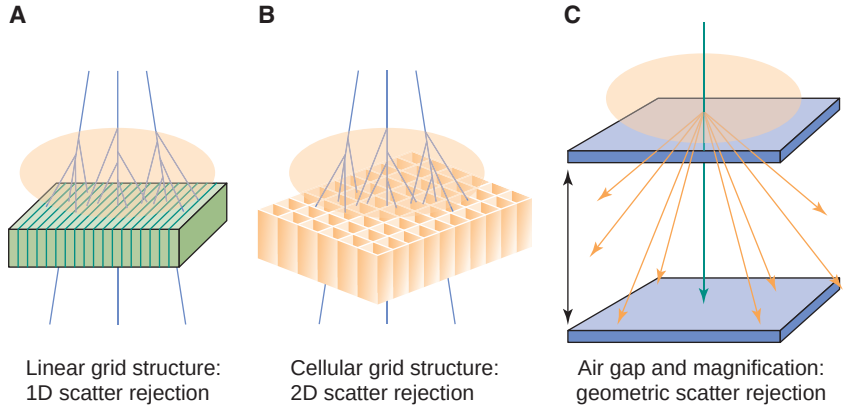


■ **FIGURE 8-16** X-ray scatter reduces the radiographic contrast of the breast image. Scatter is chiefly dependent on breast thickness and field area, and largely independent of kV in the mammography energy range (25 to 35 kV). The scatter-to-primary ratio is plotted as a function of the diameter of a semicircular field area aligned to the chest wall edge, for several breast thicknesses of 50% glandular tissue.

Scattered radiation reaching the image receptor can be greatly reduced by the use of an *antiscatter grid* or *air gap*. (Antiscatter grids are described in Chapter 7.) For contact mammography, an antiscatter grid is located between the breast and the detector. Mammography grids transmit about 60% to 70% of primary x-rays and absorb 75% to 85% of the scattered radiation. Linear focused grids with grid ratios (height of the lead septa divided by the interspace distance) of 4:1 to 5:1 are common (e.g., 1.5 mm height, 0.30-mm distance between septa, 0.016-mm septa thickness), with carbon fiber interspace materials (Fig. 8-17A). Grid frequencies (number of lead septa per cm) of 30/cm to 45/cm are typical. To avoid grid-line artifacts, the grid must oscillate over a distance of approximately 20 lines during the exposure. Excessively short exposures are the cause of most grid-line artifacts because of insufficient grid motion.

A cellular grid, made of thin copper septa, provides scatter rejection in two dimensions (Fig. 8-17B). Specifications of this design include a septal height of 2.4 mm, 0.64-mm distance between septa (3.8 ratio), a septal thickness of 0.03 mm, and 15 cells/cm. During image acquisition, a specific grid motion is necessary for complete blurring, so specific exposure time increments are necessary and are determined by AEC logic evaluation of the first 100 ms of the exposure. Because of two dimensional scatter rejection and air interspaces, the cellular grid provides a better contrast improvement factor than the linear grid.

Grids impose a dose penalty to compensate for loss of primary radiation that would otherwise contribute to SNR in the digital image or for x-ray scatter and primary radiation losses that would otherwise contribute to film optical density in film-screen mammography. As discussed in Chapter 7, the *Bucky factor* is the ratio of exposure with the grid compared to the exposure without the grid to achieve the same film optical density. For mammography grids, the Bucky factor is 2 to 3, so the breast dose is doubled or tripled, but the benefit is improvement of image contrast by up to 40%. For digital acquisitions, there is no similar "Bucky factor" definition, but the loss of signal in the digital image from primary radiation being attenuated by the grid strips will increase the exposure needed to achieve a similar SNR. Typically,



■ **FIGURE 8-17** Antiscatter devices commonly employed in mammography include (**A**) the linear grid of approximately 5:1 grid ratio and carbon fiber interspace material, (**B**) a cellular crosshatch grid structure made of copper sheet of approximately 3.8 grid ratio with air interspaces and scatter rejection in two dimensions, and (**C**) the air gap intrinsic to the magnification procedure. Note: while the illustration depicts 100% scatter rejection by the grid, approximately 15% of scattered x-rays are transmitted.

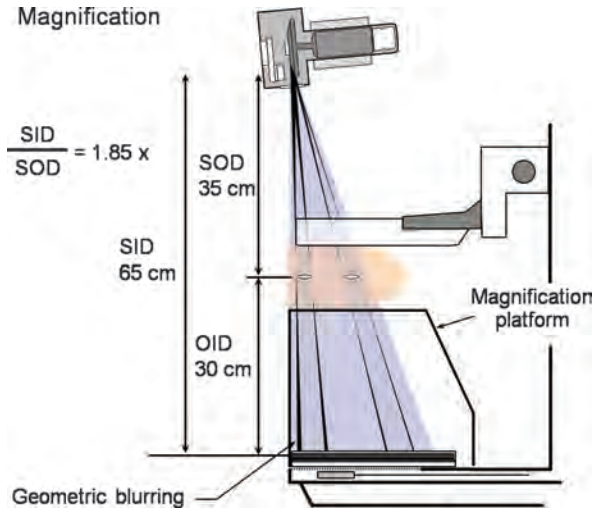
the extra radiation needed for a digital detector is less than the Bucky factor for screen-film, perhaps by one half or more. Grid performance is far from ideal, but digital acquisition and image processing can somewhat mitigate the dose penalty incurred by the use of the antiscatter grid and analog screen-film detectors.

An air gap reduces scatter by increasing the distance of the breast from the detector, so that a large fraction of scattered radiation misses the detector (Fig. 8-17C). The consequences of using an air gap, however, are that the field of view (FOV) is reduced, the magnification of the breast is increased, and the dose to the breast is increased. However, use of a sufficiently large air gap can render the antiscatter grid unnecessary, thus reducing dose to the breast by approximately the same factor.

Magnification Techniques

Geometric magnification (Chapter 7) is used to improve system resolution, typically for better visualization of microcalcifications. In magnification studies, geometric magnification is achieved by placing a breast support platform (a magnification stand) at a fixed distance above the detector, selecting the small (0.1 mm) focal spot, removing the antiscatter grid, and using a compression paddle (Fig. 8-18). Most dedicated mammographic units offer magnifications of 1.5×, 1.8×, or 2.0×. Advantages of magnification include (a) increased effective resolution of the image receptor by the magnification factor; (b) reduction of image noise relative to the objects being rendered; and (c) reduction of scattered radiation. Geometric magnification of the x-ray distribution pattern relative to the inherent unsharpness of the image receptor improves effective resolution, as long as the loss of resolution by geometric blurring is mitigated with the small 0.1-mm nominal focal spot. Since there are more x-rays per object area in a magnified image, the effective quantum noise is reduced, compared to the standard contact image. The distance between the breast support surface of the magnification stand and the image receptor reduces scattered radiation reaching the image receptor, thereby improving image contrast and reducing random noise and making an antiscatter grid unnecessary.

Magnification has several limitations, the most significant being geometric blurring caused by the finite focal spot area. Even with a small 0.1-mm focal spot, spatial resolution will be less on the cathode side of the x-ray field (toward the chest



■ **FIGURE 8-18** Geometric magnification. A support platform positions the breast closer to the source focal spot, resulting in 1.5× to 2.0× image magnification. A small focal spot (0.1-mm nominal size) reduces geometric blurring. Variation of effective focal spot size along the cathode-anode axis (large to small effective size) is accentuated with magnification and increased penumbra width. Best resolution and detail in the image exist on the anode side of the field. SID: source to image distance; SOD: source to object (midplane) distance; OID: object to image distance.

wall), where the effective focal spot length is largest. In addition, the small focal spot has a tube current limit of about 25 mA for a Mo target, so a 100-mAs technique requires 4 s exposure time. Even slight breast motion will result in image blurring, which is exacerbated by the magnification. For screen-film detectors and long exposure times, reciprocity law failure requires additional radiation exposure to achieve the desired film OD. Regarding breast dose, the elimination of the grid reduces dose by a factor of 2 to 3, but the shorter distance between the x-ray focal spot and the breast increases dose by about the same factor due to the inverse square law. Therefore, in general, the average glandular dose delivered to the breast with magnification is similar to that from contact mammography. However, the smaller irradiated FOV justifies collimating the x-ray beam to only the volume of interest, which reduces radiation scatter and breast dose.

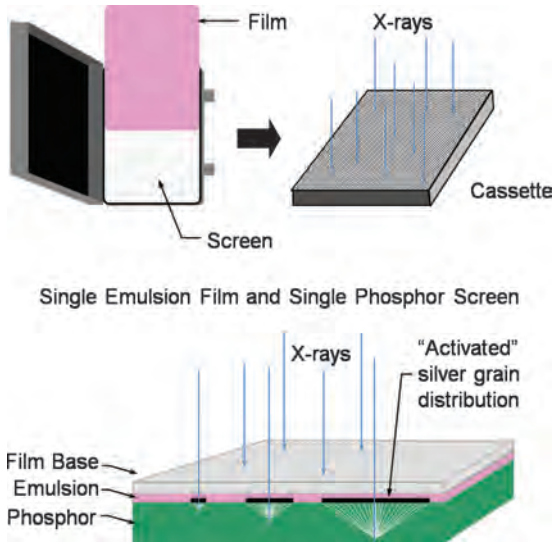
8.4 Screen-Film Cassettes and Film Processing

Screen-film detectors in mammography have been a proven technology over the last 30 years; in fact, the 2001–2005 study comparing screen-film and digital mammography detectors showed that screen-film detectors are comparable to digital detectors in terms of diagnostic accuracy as a means of screening for breast cancer. Digital mammography has been shown to be more accurate for women under the age of 50 years, women with radiographically dense breasts, and premenopausal or perimenopausal women compared to screen-film mammography (Pisano et al. 2005). While digital mammography has reduced the use of screen-film mammography, for some low-volume mammography clinics and those clinics not yet able to transition to digital technology, screen-film detectors remain a reasonable and cost-effective technology.

Design of Mammographic Screen-Film Systems

A mammographic screen-film detector is comprised of a cassette, intensifying screen or screens, and a light-sensitive film. Most cassettes are made of low-attenuation carbon fiber and have a single high-definition phosphor screen used in conjunction with a single emulsion film. Terbium-activated gadolinium oxysulfide (Gd_2O_2S:Tb) is the most commonly used screen phosphor. This scintillator emits green light, requiring a green-sensitive film emulsion. Intensifying screen-film speeds and spatial resolution are determined by screen phosphor particle size, number of phosphor particles per volume, light-absorbing dyes in the phosphor matrix, and phosphor layer thickness. Mammography screen-film systems have their own speed ratings and are classified as regular (100 or par speed) and medium (150 to 190 speed), whereby the 100-speed screen-film system requires 120 to 150 μGy incident air kerma to achieve the desired film OD. For comparison, a conventional "400-speed" screen-film cassette for general diagnostic imaging requires about 5-μGy air kerma.

The intensifying screen is positioned in the *back* of the cassette so that x-rays travel through the cassette cover and the film before interacting with the phosphor. Exponential attenuation in the phosphor results in a greater fraction of x-ray interactions near the phosphor surface closest to the film emulsion. At shallow depths in the phosphor, absorbed x-rays produce light distributions that have minimal light spread prior to interacting with the film, thus preserving spatial resolution. X-ray absorption occurring at greater depth in the screen produces a broader light distribution and reduces spatial resolution (Fig. 8-19), but the number of interactions



■ **FIGURE 8-19** Nearly all screen-film systems used in mammography employ a single phosphor screen with a single emulsion film and are "front-loaded" (x-rays pass through the film to the screen). Light spread varies with the depth of x-ray absorption, as depicted in the lower diagram. Most x-rays interact in the layers of the screen closest to the film, preserving spatial resolution because of reduced spread of the light distribution.

decreases exponentially with depth. For a "100-speed" mammography screen-film cassette, the limiting spatial resolution is 15 to 20 lp/mm.

Selection of a screen-film combination requires consideration of measures such as spatial resolution and radiographic speed, as well as factors such as screen longevity, cassette design, film base color, film contrast characteristics, radiologist's preference, and cost. In today's market, there is no single best choice; hence, most screen-film receptors, when used as recommended with particular attention to film processing, provide excellent image quality for mammography imaging.

Film Recording and Film Processing

Mammography films are sensitized by x-ray–induced light from the phosphor screen. Light produces latent image centers on microscopic light-sensitive silver halide grains in the emulsion layered on a film substrate. (See Chapter 7 for more information.) Intensity variations in the light produce corresponding variations in the number of sensitized grains per area. Chemical processing in a bath of developer solution renders the invisible latent image into a visible image by reducing the sensitized grains to elemental silver. Unsensitized grains in the film emulsion are not reduced and are dissolved and washed away by the fixer solution. Subsequent rinsing and drying complete the film processing. Film is thus a two-dimensional recording of the incident light patterns encoded by silver grains on the substrate that produce an image with varying light intensities when the film is placed on a lightbox. Film opacity is the degree of the obstruction of light transmission (the inverse of transmittance), and optical density (OD), equal to the base ten logarithm of the opacity, is used as the metric of film response, as the eye responds linearly to variations in OD (e.g., an OD = 2 appears to be twice as "dark" as an OD = 1, although 10 times more light is attenuated with an OD = 2 compared to OD = 1). The useful OD levels on a mammography film range from about 0.2 up to about 3.

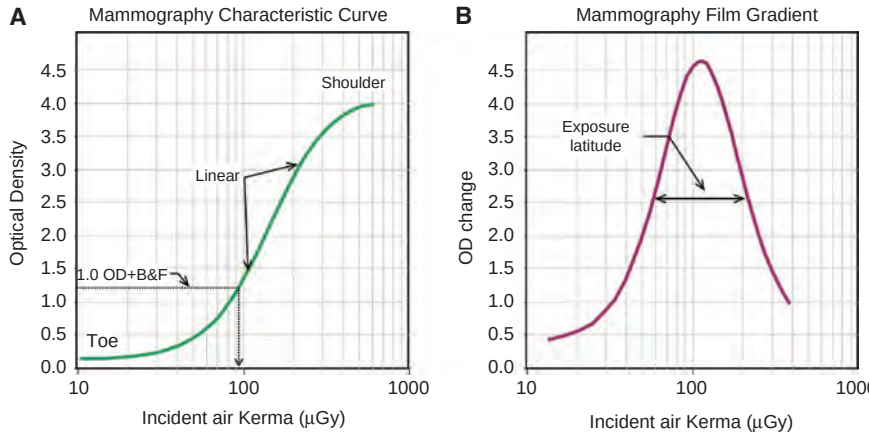
Film processing is a critical step in the mammographic imaging chain for screen-film detectors and is performed by automatic film processors. Film characteristics such as film speed (an indication of the amount of incident radiation required to achieve a specified OD on the film) and film contrast (the rate of change of OD for a known difference in incident radiation) are consistently achieved by following the manufacturer's recommendations for developer formulation, development time,

developer temperature, developer replenishment rate, and processor maintenance. If the developer temperature is too low, the film speed and film contrast will be reduced, requiring a compensatory increase in radiation dose. If the developer temperature is too high or immersion time too long, an increase in film speed occurs, permitting a lower dose; however, the film contrast is likely to be reduced because film fog and quantum mottle are increased. Stability of the developer may also be compromised at higher-than-recommended temperatures. The manufacturer's guidelines for optimal processor settings and chemistry specifications should be followed. Also, because mammographic films must be retained for years, fixer retention on the film must be measured when evaluating the processor. If film washing is incomplete and fixer is retained, the films will oxidize and image quality will deteriorate over time.

Film Response, Sensitometry, and Quality Control

A film characteristic curve, also referred to as an *H* and *D curve*, represents the relationship between the incident x-ray exposure (proportional to light intensity on the film) and the resulting OD in the processed film. As shown in Figure 8-20A, a film characteristic curve is a graph of the OD as a function of the logarithm of the relative exposure and has three major sections: (a) the *toe* has a low OD that changes very little with incident exposure up to a threshold; (b) the *linear section* represents increasing OD with increasing incident exposure, with subject contrast mapped as optical densities that render the radiographic contrast; and (c) the *shoulder* represents saturation of the film, where increases in incident exposure do not significantly increase the OD. A film's characteristic curve is determined by both the design of the film and the film processing. Another way of displaying the film's response is the gradient curve, shown in Figure 8-20B. The gradient curve is a graph of the change of OD per unit change in the incident exposure, as a function of the logarithm of the relative exposure. The gradient is the rate of change of the characteristic curve, that is, the film contrast. The Film contrast changes with incident exposure and has a value near the maximum only over a narrow range of incident exposures.

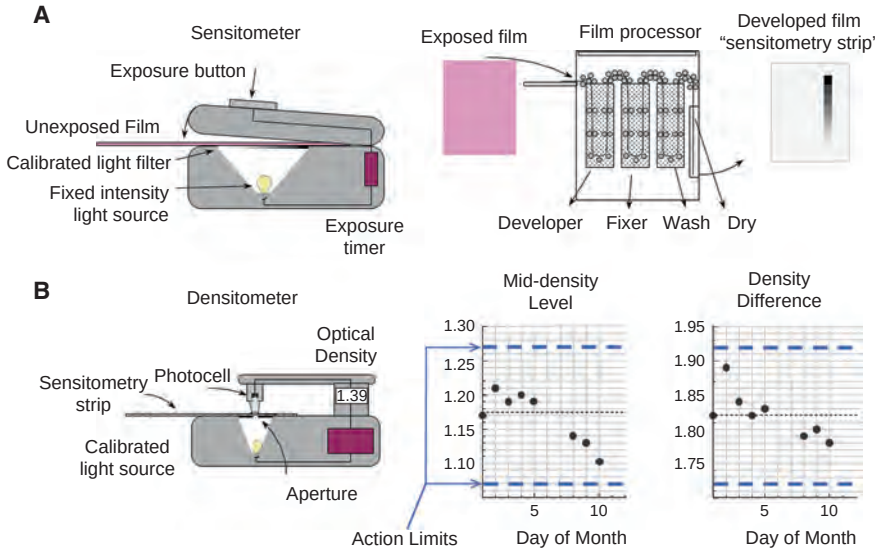
Because of the high sensitivity of mammography film quality to slight changes in processor performance, routine monitoring of film contrast, speed, and base plus fog OD is important. A film-processor QC program is required by the MQSA



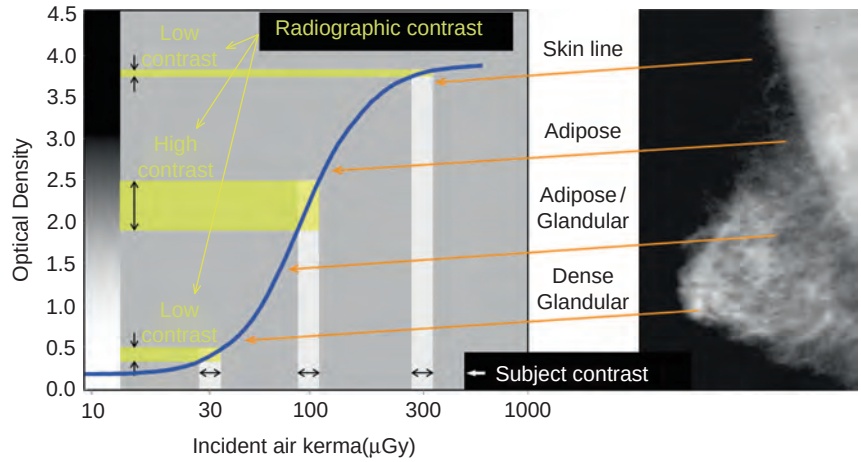
■ **FIGURE 8-20** **A**. The film characteristic curve is a graph of the OD as a function of the logarithm of incident air kerma, μGy (or exposure, mR). **B**. The gradient curve (the rate of change of the characteristic curve) indicates the contrast as a function of the logarithm of air kerma (exposure). The maximal value of the gradient curve is achieved for an exposure of about 120 μGy (14 mR), which is OD of about 2.0 for this exposure. Thus, the maximal contrast (change in OD for a change in exposure) is achieved at an OD of about 2.0.

regulations, and daily sensitometric tests must be performed before the first clinical images to verify acceptable performance. This requires the use of a sensitometer and a densitometer. A sensitometer is a device that emulates a range of incident radiation exposures by using a constant light source and calibrated optical attenuation steps to expose a mammographic film to known relative light intensities. Developing the film produces a range of numbered OD steps known as a *film sensitometric strip*. A densitometer is a device that measures the light transmission through a small area of the film and calculates the OD. Additionally, a thermometer is used to measure the temperature of the developer solution in the processor and monitoring chart are tools used to complete the processor evaluation. A chart is used to record and monitor the quality control measurements.

A box of film is reserved for QC testing; using films from a single box eliminates variations among different lots of film from the manufacturer. Daily, in the darkroom prior to the first case of the day, a film is exposed using a calibrated sensitometer and is then developed (Fig. 8-21A). On the processed film, optical densities on the sensometric strip are measured with a film densitometer (Fig. 8-21B). Data, which are plotted on a processor QC chart, are the base plus fog OD (measured at an area on the film that was not subjected to light exposure), a mid-density speed index step, and a density difference (an index of contrast) between two steps. The mid-density OD step and the two steps for calculating the density difference are selected from 5 day average ODs when the QC program is established. The mid-density step is the one whose OD is closest to, but does not exceed 1.2; the density difference steps are those whose ODs are closest to but not exceeding 0.45 and closest to 2.20. Action limits, determined from the initial 5-day averages, define the range of acceptable processor performance; they are *base + fog density* not exceeding +0.03, *middensity* ± 0.15 OD, and *density difference* ± 0.15 OD. If the test results fall outside of these limits, corrective action must be taken prior to patient imaging.



■ **FIGURE 8-21** **A**. Film-processor testing requires a calibrated sensitometer to "flash" an unexposed film to a specified range of light intensities using an optical step wedge. The film is then developed. **B**. The corresponding optical densities on the processed film are measured with a densitometer, and Base + Fog (the lowest OD value), the mid-density level step (a measurement indicating relative speed), and the density difference value, calculated by the subtraction of a low OD step from a high OD step (indicating the relative contrast response), are plotted to verify processor performance before patients are imaged.



■ **FIGURE 8-22** Equal x-ray subject contrast (range of transmitted relative exposures) generated in the breast are manifested with variable contrast in the film image, resulting from the nonlinear characteristic curve. Areas in the breast that have high transmitted exposure, such as the skin line, and low transmitted exposure, such as glandular tissue, will have reduced radiographic contrast. Only over a small range of exposures is the radiographic contrast high.

Variability of Screen-Film Contrast Response

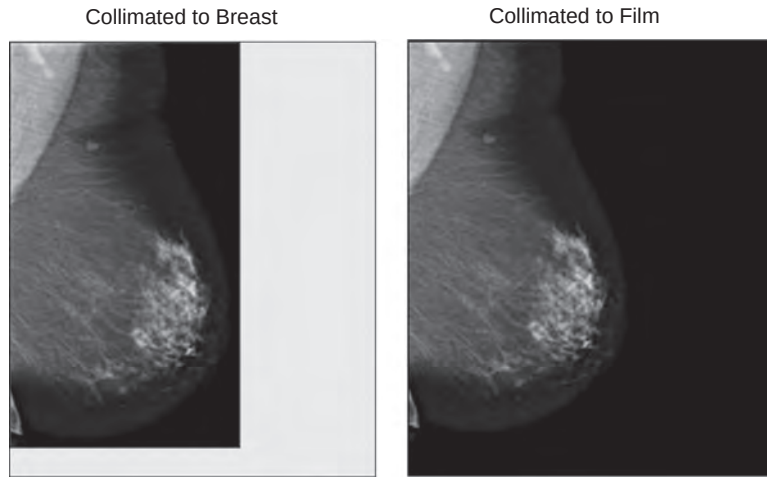
Radiographic contrast (i.e., OD differences on the developed film) results from variations in subject contrast (x-ray transmission differences due to anatomy) translated into these OD differences according to the film's characteristic curve. Therefore, the radiographic contrast varies depending upon the incident exposure to each area on the film-screen cassette. The best recording of subject contrast occurs on areas of the film where the exposure corresponds to the steepest part of the film characteristic curve. Significant reduction of contrast occurs on parts of the film where the exposures correspond to the toe and shoulder segments of the curve. Unfortunately, not all important anatomical information is rendered with optimal contrast, particularly in dense glandular breast tissues and at the skin line, as illustrated in Figure 8-22.

Film Viewing Conditions

Viewing conditions are very important so that the information on the film is visualized well by the radiologist. Subtle lesions on the mammographic film may be missed if viewing conditions are suboptimal. Since mammography films are exposed to high optical densities to achieve high contrast, view boxes in the mammography reading area must provide high *luminance* (a measure of brightness as described in Chapter 5). Mammography view boxes should have minimum luminances of at least 3,000 cd/m^2, and luminances exceeding 6,000 cd/m^2 are common. (Note: the luminance of 1 cd/m^2 is approximately the luminous intensity of the sky at the western horizon on a clear day, 15 min after sunset.) In comparison, the luminance of a view box in diagnostic radiology is typically about 1,500 cd/m^2.

A high-intensity "bright" light should be available in the reading room to penetrate high OD regions of the film, such as the skin line and the nipple area. Furthermore, the use of a magnifying glass improves the visibility of fine detail in the image, such as microcalcifications.

Film masking (blocking clear portions of the film and areas of the view box that are not covered by film to prevent extremely bright light from degrading the



■ **FIGURE 8-23** Optimal viewing of film mammograms requires that the entire area of the film be exposed to radiation to darken the background surrounding the breast image; otherwise, light cannot easily be masked when viewing the films on a viewbox. For digital acquisitions, the background area location in each image is determined, and "pixel padding" is performed so that the background is dark, even when digitally inverting contrast.

radiologist's visual adaptation to the range of luminances displayed by the images) is essential for preserving perceived radiographic contrast. The film must be fully darkened in background areas around the breast (Fig. 8-23) by collimating the x-ray beam to the full area of the detector (see Section 8.1: "Collimation"). Most dedicated mammography view boxes have adjustable shutters for masking to the film size.

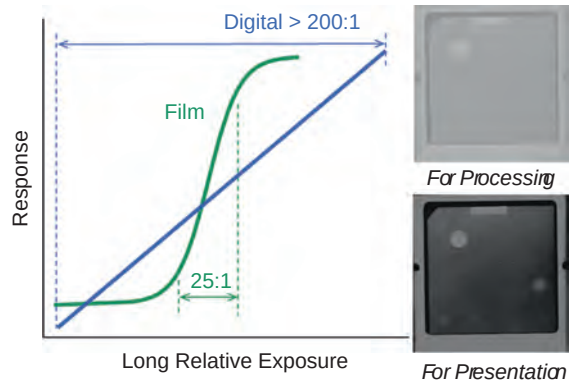
The ambient light intensity (as measured by the *illuminance*) in a mammography reading room must be reduced to eliminate reflections from the film and to improve perceived radiographic contrast. Illuminance is the luminous flux incident upon a surface per unit area, measured in lux or lumens/m^2. For example, the full moon in a clear night sky produces about 1 lux, whereas normal room lighting generates 100 to 1,000 lux. Subdued room lighting providing an illuminance of about 20-40 lux is acceptable for the viewing and interpretation of mammograms.

8.5 Digital Mammography

Digital acquisition devices became available in mammography in the early 1990s in the form of small field-of-view digital biopsy systems. In early 2000, a full-field digital mammography system was first approved by the FDA; it had a thin-film-transistor (TFT) indirect detection flat panel array with an 18 × 23-cm active area. Since that time, several digital mammography systems have been approved by the FDA in the United States and, in 2010, 70% of all clinical mammography machines in the United States were digital systems.

Full-Field Digital Mammography

There are compelling advantages to digital mammography, the most important of which is the ability to overcome the exposure latitude limitations of screen-film detectors and produce better image quality at lower doses. Other reasons include

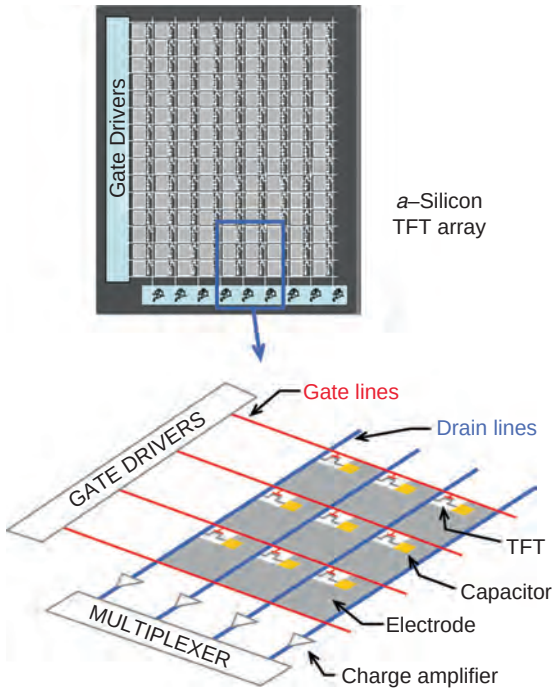


■ **FIGURE 8-24** A comparison of the response of a screen-film and digital radiography detector to variations in relative exposure shows the large differences in exposure latitude and image response. The *For Processing* image (DICOM term for a raw image that has been corrected for flaws and detector non-uniformities) is linearly related to incident exposure; the *For Presentation* image is postprocessed by contrast and spatial resolution enhancements for viewing.

increased technologist productivity through rapid acquisition and evaluation of digital images; reduced patient waiting time, particularly for diagnostic studies; and improved image quality and conspicuity of lesions.

Screen-film detectors for mammography are excellent image capture devices, and the film serves as both the display device and the storage media, but the major drawbacks are the limited exposure dynamic range and narrow latitude, which are necessary to achieve high radiographic contrast (Fig. 8-24). A highly glandular breast can produce exposure latitude that exceeds 200:1, causing underexposed film areas corresponding to the glandular tissue, and overexposed film areas corresponding to the skin line and thinner parts of the breast. Digital flat panel and digital cassette (CR) detectors have exposure dynamic ranges greater than 1,000:1, which result in images showing little contrast when the entire ranges of pixel values are displayed. The full exposure data is contained in the the raw, *For Processing* image shown in Figure 8-24. With mammography-specific image postprocessing, high image contrast can be rendered over all exposure levels (e.g., from the high detector exposure at the skin line to the low detector exposure behind dense glandular areas of the breast), as shown in the processed *For Presentation* image. Postprocessing image enhancement is a key element in producing an optimized digital mammogram.

Flat panel detector arrays, which became clinically available in the mid to late 1990s for general radiography (see Chapter 7), are the major detector technologies used in digital mammography. An active matrix, thin film transistor (TFT) array collects the local signal (electric charge) generated during the x-ray exposure, absorption, and conversion process; stores the charge in a capacitor attached to each detector element; and actively reads the array immediately afterward to produce the image (Fig. 8-25). Key components in each detector element include a transistor (which serves as an "on-off" switch), a charge collection electrode, and a storage capacitor. During image acquisition, the transistor, controlled by the gate line, is in the "off" position, and the charge collected by the electrode from incident x-rays is stored in the capacitor. Immediately after the exposure, readout occurs by activating each gate line, one row at a time, which turns on the transistors along each row, allowing charge transfer via the drain lines along each column to a series of charge amplifiers and digitizers. The digital image is then stored in computer memory. In some "fast" flat panel designs, readout of the whole array is performed in hundreds of milliseconds, allowing near real-time acquisition of data for applications such as digital tomosynthesis.

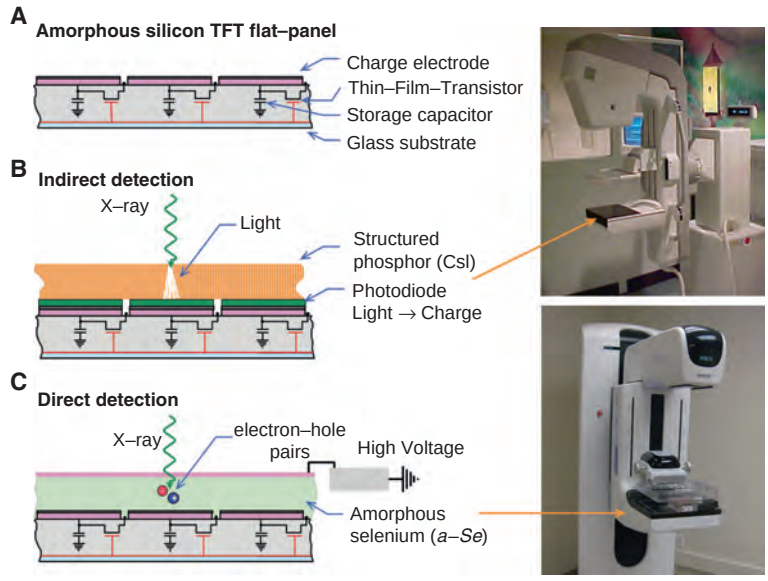


■ **FIGURE 8-25** The flat panel array is a two-dimensional matrix of individual detector elements lithographed on an amorphous silicon substrate. Each detector element is comprised of an electronic switch (the TFT), a charge collection electrode, and a storage capacitor. The gate and drain lines provide the mechanism to extract the locally stored charge by activating gate lines row by row, and collecting charge down each column. Charge amplifier and digitizers convert the signals to corresponding digital values for transfer to the digital image matrix.

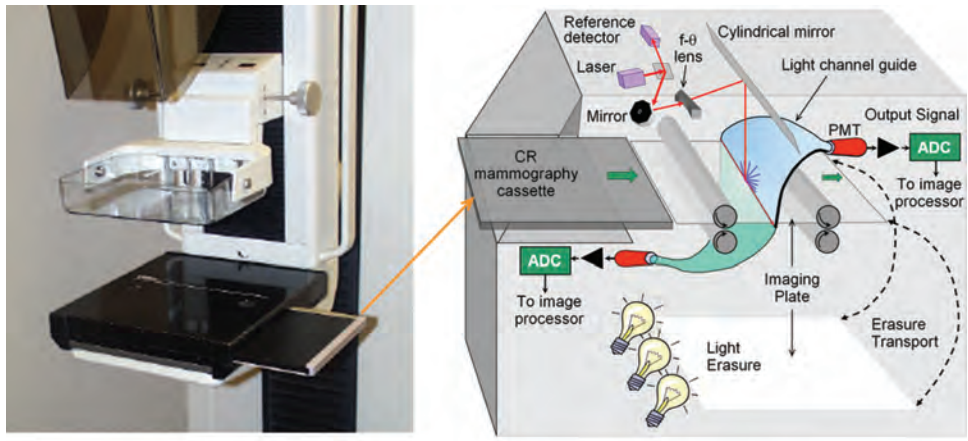
Common to all flat panel TFT arrays is the amorphous silicon circuit layer (Fig. 8-26A). At present, approved digital mammography systems are based on four technologies. The first is an indirect x-ray conversion TFT flat panel array receptor with 100-μm sampling pitch and 18 × 23-cm or 24 × 31-cm FOV (Fig. 8-26B). A structured cesium iodide (CsI) phosphor converts x-rays into light; the light is emitted onto a photodiode in each detector element, from which the charge is generated and stored in the local capacitor. *Indirect conversion* describes absorption of x-rays in the CsI, the production of secondary light photons directed to a photodiode, and the generation of the charge, which is stored on the storage capacitor in that detector element.

A second technology is based on a direct x-ray conversion TFT detector with 70-μm sampling pitch and 24 × 29-cm FOV (Fig. 8-26C). A semiconductor x-ray converter, Se, is uniformly deposited between two electrodes, with a large voltage placed across the Se layer. Incident x-rays absorbed by the Se directly produce electron-hole pairs; the applied voltage causes the electrons to travel to one electrode and the holes to the opposite electrode. The local storage capacitor captures the charge from the collection electrode. *Direct conversion* refers to the direct generation of charge by x-rays within the photoconductor and capture by the electrode without intermediate signals.

A third technology approved for digital mammography is a cassette-based, dual-side readout CR photostimulable storage phosphor (PSP) imaging plate detector and reader system. Cassettes of 18 × 24 cm and 24 × 30 cm are used in place of screen-film cassettes (Fig. 8-27), whereby x-rays absorbed in the phosphor raise electrons to higher energy levels in the PSP material, where a fraction of them are caught in semi-stable electron traps. An exposed imaging plate is subsequently processed by a reader system that scans the plate with a laser beam with an effective spot size of 50 microns. The laser beam provides energy to free the stored excited electrons from the traps; these electrons fall to a lower energy state with the emission of light, a process called "stimulated luminescence." Some of the light reaches a photomultiplier tube (PMT), which produces and amplifies an electrical current



■ **FIGURE 8-26** **A**. TFT flat panel arrays have a common underlying amorphous silicon structure, but there are two types of TFT detectors, determined by the conversion of the x-rays into electrical signals. **B**. "Indirect x-ray conversion" TFT arrays have an additional photodiode layer placed on the charge collection electrode of each detector element. The photodiodes are optically coupled to a layer of a phosphor, such as CsI, that produces light when irradiated by x-rays. The light produced by x-ray absorption in the phosphor in turn creates mobile electric charge in the photodiode. The charge is collected by the electrode and stored in the capacitor. **C**. A "Direct x-ray conversion" TFT array has a semiconductor layer of approximately 0.5 mm thickness between the surface electrode extending over the detector area and each detector element electrode, under a potential difference of approximately 10 V/μm. As hole-electron pairs are created in the semiconductor layer by x-rays, the electrons and holes migrate to the positive and negative electrodes, with minimal lateral spread. A proportional charge is stored on the capacitor. The outward appearances of the indirect and direct detectors are similar, shown by the pictures on the right.



Cassette-based digital detector and "dual-side" readout of imaging plate

■ **FIGURE 8-27** The CR digital mammography detector uses a cassette that contains a PSP imaging plate, compatible with screen-film cassette sizes. Workflow is similar to screen film, but the AEC settings require independent calibration. When exposed to x-rays, the storage phosphor contains a "latent image" of trapped, excited electrons. Subsequent imaging plate processing uses dual-side readout (Fig. 8-27, **right**) by a scanning 0.050-mm laser beam, plate translation, and approximately 60-s readout time. Stimulated luminescence produces a light signal at each point on the imaging plate that is detected and digitized to create the digital image. Erasure of the imaging plate after the readout eliminates residual signals prior to next use.

proportional to the light intensity. The signal from the PMT is digitized to provide pixel values of the digital image. Typical turnaround time of an imaging plate is on the order of 60 to 90 s, similar to the batch mode processing of screen-film images.

The fourth technology approved for digital mammography uses a narrow 22 cm long charge-coupled-device (CCD)-detector array, optically coupled to a CsI phosphor x-ray converter, to capture x-ray signals in a highly collimated x-ray beam configuration (a "slot"). The moving collimator slot sweeps a narrow x-ray beam across the breast. The detector moves across the image plane to stay aligned with the scanning-slot x-ray beam. The scan acquires data in the medial to lateral directions in about 5 s, with 50-μm sampling over a 22 × 29-cm FOV. A major advantage of the scanning-slot fan-beam system is that very little scattered radiation from the breast reaches the detector. This technology is not being manufactured at the present time.

All current full-field digital mammography detectors have similar attributes with respect to exposure latitude. In terms of spatial resolution, the direct conversion TFT detector achieves the best intrinsic spatial resolution due to active charge collection and minimal signal spread in the semiconductor material, the signal being acquired onto each 70 μm × 70 μm detector element (dexel). The indirect conversion TFT detector uses a CsI(Tl)-structured phosphor to reduce lateral spread of the light photons onto a 100 μm × 100 μm dexel, and the CR system uses a 50-μm scanning laser beam and plate translation speed to maintain an effective 50 μm × 50 μm dexel. In terms of detective quantum efficiency (DQE—defined in Chapter 4), the indirect TFT array has better performance at low to intermediate spatial frequencies, and the direct TFT array has better performance at higher spatial frequencies. For cassette-based CR mammography detectors, the image quality and SNR at a given incident radiation exposure to the detector is lower than that of the TFT array detectors. In general, mammography technique factors and the corresponding average glandular doses are lowest for the indirect TFT detectors and highest for the cassette-based CR detectors. In most situations (with appropriate calibration of the AEC), use of digital detector systems will result in lower doses than screen-film detectors, particularly for thick, dense breast tissues.

Due to high-spatial resolution requirements, digital mammography images are quite large. A digital mammography system with a 0.07-mm pixel, 2 bytes per pixel, and a 24 × 29-cm active detector area, produces a 27-MB image. One screening study (4 images) will therefore contain 108 MB of data and, if 3 years of prior images are reviewed, 432 MB of data will be required per patient reading session. The network overhead and storage required for the picture archiving and communications system are very large and even larger with the increased number of images in a diagnostic exam. In addition, if *For Processing* images are stored, another factor of two increase is incurred. Typical digital image sizes for mammography systems are listed in Table 8-6.

TABLE 8-6 DIGITAL IMAGE SIZE FOR MAMMOGRAPHY SCREENING EXAMS

DETECTOR TYPE	FOV (cm)	PIXEL SIZE (mm)	IMAGE SIZE (MB)	EXAM SIZE (MB)	+3 Y PRIORS (MB)
Indirect TFT	19 × 23	0.10	9	35	140
Indirect TFT	24 × 31	0.10	15	60	240
Direct TFT	18 × 24	0.07	18	70	280
Direct TFT	24 × 29	0.07	27	108	432
CR	18 × 24	0.05	32	128	512
CR	24 × 30	0.05	50	200	800

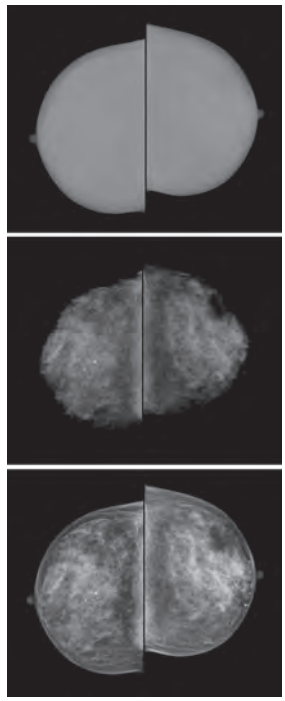
Currently, the recommendation regarding mammography image compression to reduce file size is to not use lossy compression algorithms—that is, an exact (lossless) image must be reproduced during the decompression step. For most digital mammograms, there is a significant background area that is "pixel-padded" with a constant value number in the *For Presentation* image. On average, a compression factor of 3 to 5 can be achieved using lossless compression, depending on the fraction of the image matrix area comprising the breast. FDA regulations require that facilities maintain mammography films, digital images, and reports in a permanent medical record of the patient for a period of not less than 5 years and not less than 10 years if no additional mammograms of the patient are performed at the facility. A longer retention period may be mandated by state or local law. Many considerations are described in the ACR practice guideline, "Determinants of Image Quality in Digital Mammography" available on the ACR website, http://www.acr.org.

Processing and Viewing Digital Mammograms

Even the best digital detectors have malfunctioning detector elements, column and line defects, and variation in gain across the detector area. Without corrections, these artifacts would result in unacceptable image quality and interfere with the diagnostic interpretation. Manufacturers use "preprocessing" algorithms to correct these defects. First, the locations of inoperative detector elements and line defects are determined, and these data are interpolated from nearest-neighbor pixels. Next, a uniform beam of x-rays exposes the detector and generates an image, showing the variable gain from the detector submodule electronics. Multiple images are averaged together to create a *flat-field* correction image, which is normalized and inverted as an inverse-gain map. When applied to an uncorrected image on a pixel-by-pixel basis, the gain variations are canceled, and the corrected "raw" image is defect-free (see Chapter 4). In practice, flat-field gain maps for various acquisition conditions are measured, stored on the system, and applied to each clinical image that is acquired under similar conditions.

The output *For Processing* image has wide latitude, digital signals that are proportional to the detector exposure or the logarithm of exposure, and very low contrast (Fig. 8-28A). While not useful for diagnosis by the radiologist, this is the preferred image for *Computer-Assisted Detection* (CADe) programs, which use computer algorithms to identify image features that may indicate the presence of breast cancer. Contrast enhancement of the *For Processing* image is shown in Figure 8-28B, which looks similar to a screen-film image. Nonlinear postprocessing algorithms applied to the *For Processing* image data identify the skin line of the mammogram, modify the local-area pixel values, and equalize the response in the high-signal, low-attenuation skin area to the other areas of the breast. Contrast and spatial resolution enhancements are then applied to create the *For Presentation* image for viewing and diagnosis. Unlike film radiographic contrast, which is variable in the areas corresponding to the thinnest and thickest parts of the breast image (see Fig. 8-22), image contrast is equalized across the entire the digital image (Fig. 8-28C). Each manufacturer has proprietary methods of image processing; the radiologist must to adopt a particular "look" for viewing current and prior digital mammograms for consistency. The *For Processing* images should be saved in addition to the *For Presentation* images as part of a patient's examination if CAD algorithms will be used in possible subsequent applications, or if other image processing algorithms are to be applied, since *For Presentation* images cannot be "unprocessed." The disadvantage to saving the *For Processing* images is the extra storage required.

It is strongly recommended that diagnostic interpretation of digital mammograms be performed on display monitors approved by the FDA for mammography having a minimum of 5 million pixels (5 MP) and a calibrated, sustained maximal luminance



■ **FIGURE 8-28** **A**. The *For Processing* images of the right and left cranial-caudal breast projection images have detector and gain map corrections applied with unity radiographic contrast. **B**. Linear contrast and brightness adjustments are applied. Appearance is similar to a screen-film image. **C**. Nonlinear processing identifies the breast skin line and densest parts of the image, to equalize the response of the processed image, with contrast and spatial resolution enhancement.

of at least 450 cd/m^2 (per ACR guidelines for digital mammography), but preferably at 600 cd/m^2. Monitors of lower luminance make reading conditions more susceptible to poor ambient lighting and illuminance conditions. A typical workstation has two large-format portrait-oriented LCD monitors (each 54 cm diagonal, 34 cm width, 42 cm height) with 2,560 × 2,048 pixels at a pixel pitch of approximately 0.165 mm and a contrast ratio (ratio of brightest to darkest luminance) greater than 350:1. The optimal viewing distance for the resolution provided by the 0.165-mm pixel is about 50 cm (slightly less than arm's length). A lower resolution, smaller format "navigation" monitor is used for patient lists and operating system access. All monitors should be calibrated to conform to the DICOM Grayscale Standard Display Function, as described in Chapter 5, to ensure image contrast is consistently and optimally displayed.

To optimize the workflow of the radiologist, mammography workstations should provide automated *mammography-specific hanging protocols* that present the images in a useful and sequential way for review and diagnosis according to view, laterality, magnification, and prior examination comparisons. Demographic overlays (patient name, technique factors, and details of the exam) should be mapped in areas outside of the breast anatomy. It is important to display images in *true size* at one point in the diagnostic review (as close as possible to a magnification of 1, *independent of the detector pixel dimension* so that measurements can be made directly on the monitor if necessary) and to effectively handle images of different size from different manufacturers to appear the same size on the workstation when direct comparisons are made. Also, viewing of *all* mammograms at one image pixel to one monitor pixel ensures that the monitor does not limit the intrinsic resolution of the image. This requires a "zoom and pan" procedure for each image. Sometimes, radiologists will use a magnifying glass on a displayed digital mammogram to overcome limitations of the human visual system acuity; this is reasonable as long as the image (or portion of) is displayed at full resolution. However, a digital magnification by pixel replication and interpolation to enlarge the image directly on the monitor is preferable. Issues

confronting digital mammography workflow and hanging protocols are addressed in the Integrating the Healthcare Enterprise Mammography Image Integration Profile*. Even after solving many of the display concerns, image appearance may be considerably different between the manufacturers' proprietary processing algorithms, which can make comparisons challenging.

An alternative to "soft-copy" reading of digital mammograms is the use of an FDA-approved laser printer to record all digital studies on film, an unlikely scenario. However, there is FDA requirement to have the capability of printing film images from digital mammograms at each facility to be able to provide hard-copy images to patients upon request. QC procedures must be performed on the printer periodically, as described in the printer manufacturer's QC manual for digital mammography printers.

Computer-Aided Detection and Computer-Assisted Diagnosis Systems

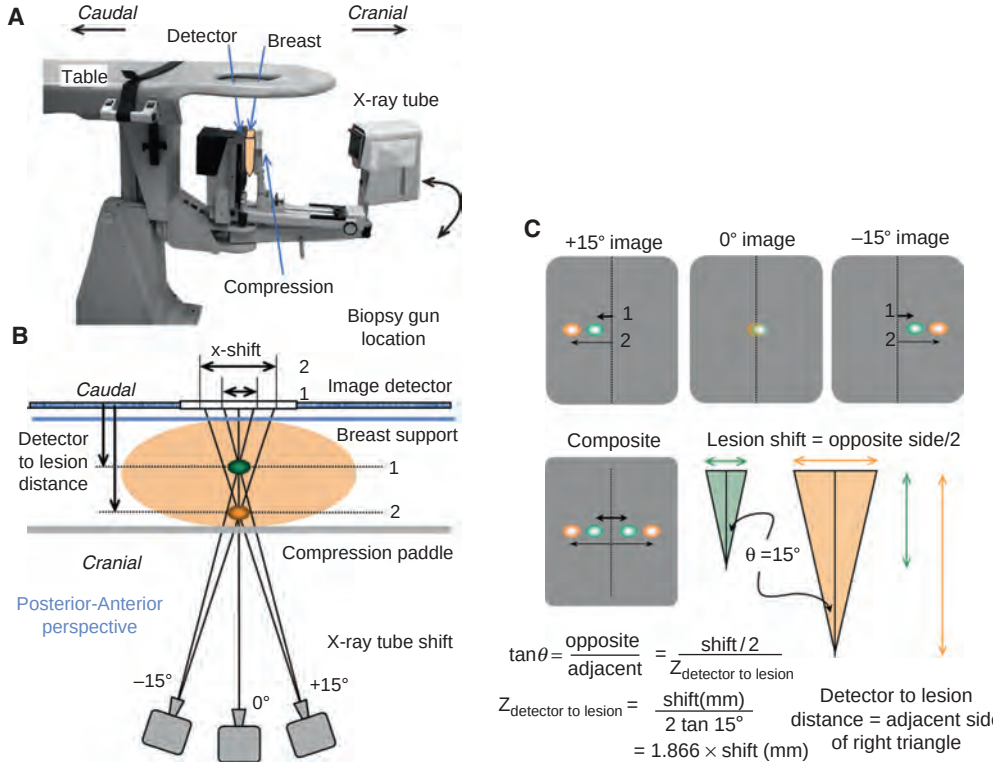
A computer-aided detection (CADe) system is a computer-based set of algorithms that incorporates pattern recognition and uses sophisticated matching and similarity rules to flag possible findings on a digital mammographic image, which may be a digitized film mammogram. The computer software searches for abnormal areas of density, mass, calcifications, and/or structural patterns (e.g., spiculated masses, architectural distortion) that may indicate the presence of cancer. The CADe system then marks these areas on the images, alerting the radiologist for further analysis. For digitally acquired mammograms, the CADe algorithms require the use of linear, *For Processing* images for analysis, as the algorithms are trained on these images and evaluation of processed images would result in lower performance in terms of sensitivity and specificity.

Computer-assisted diagnosis (CADx) devices include software that provides information beyond identifying suspicious findings; this additional information includes an assessment of identified features in terms of the likelihood of the presence or absence of disease, or disease type. For instance, a CADx system for mammography will not only identify clusters of microcalcifications and masses on digital mammograms but also provide a probability score of the disease or malignancy to the radiologist for each potential lesion. Currently, CADe and CADx systems are widely used by radiologists who interpret mammography studies in addition to other duties. The fraction of False-positive marks (marks placed on image features that do not correspond to malignancy) can be high and is one of the downsides of using these systems. Many expert mammography radiologists avoid using CAD for this reason.

Stereotactic Breast Biopsy

Stereotactic breast biopsy systems provide the capability to localize breast lesions in three dimensions and physically sample them with a targeted biopsy instrument. Generally, these systems are used for targeting microcalcifications associated with lesions, whereas localization and biopsy of masses and areas of architectural distortion are performed either with ultrasound guidance or MRI. Figure 8-29A shows a prone-geometry stereotactic imaging system. The image receptors of these systems are CCD cameras coupled to x-ray phosphor screens of 5 cm × 5-cm FOV by either lens optics or fiber optics. A 1,000 × 1,000 detector element matrix, with an

*http://www.ihe.net/Resources/handbook.cfm)



■ **FIGURE 8-29** **A**. A stereotactic biopsy system with a prone positioning system is shown. The patient's pendulant breast is positioned on the detector platform and compressed by an open-access paddle. The x-ray tube pivots about a fulcrum point to acquire digital images at +15-, and –15-degree projections within the 5 × 5-cm detector FOV. **B**. Breast lesions positioned more cranially (toward the tube) will have a greater shift in the image pair than lesions closer to the detector. **C**. Trigonometric relationships allow the determination of lesion depth from the measured shift distance in the images. A frame of reference calibration between the images on the computer and the biopsy gun placed on rails between the x-ray tube and breast allows a trajectory calculation and needle orientation under computer control.

effective detector element size of 50 μm (0.05 mm), which is equivalent to10 lp/mm, is deemed sufficient to visualize microcalcifications. Image acquisition is performed by integrating light from x-ray exposure of the phosphor screen on the CCD photodetector, reading the detector element array one line at a time after the exposure, amplifying the electronic signal, and digitizing the output.

The breast is aligned and compressed by an open-area paddle. After verification with a scout image, two images of the breast are acquired, at +15 and –15 degrees from the normal, causing the lesion projections to shift on the detector, as shown in Figure 8-29B. The shift distance is measured (Fig. 8-29C) from trigonometry relationships, the opposite sides of two right triangles, and the common adjacent side, each with a 15-degree angle (30 degrees total). The adjacent side represents the lesion distance from the detector, and since Opposite/Adjacent = tan(θ = 15 degrees), the lesion distance from the detector (mm) is calculated as 1.866 × shift (mm). Trajectory information is determined by the position of the marked points in the images correlated with the frame of reference of the biopsy device, which is positioned and activated to capture the tissue samples.

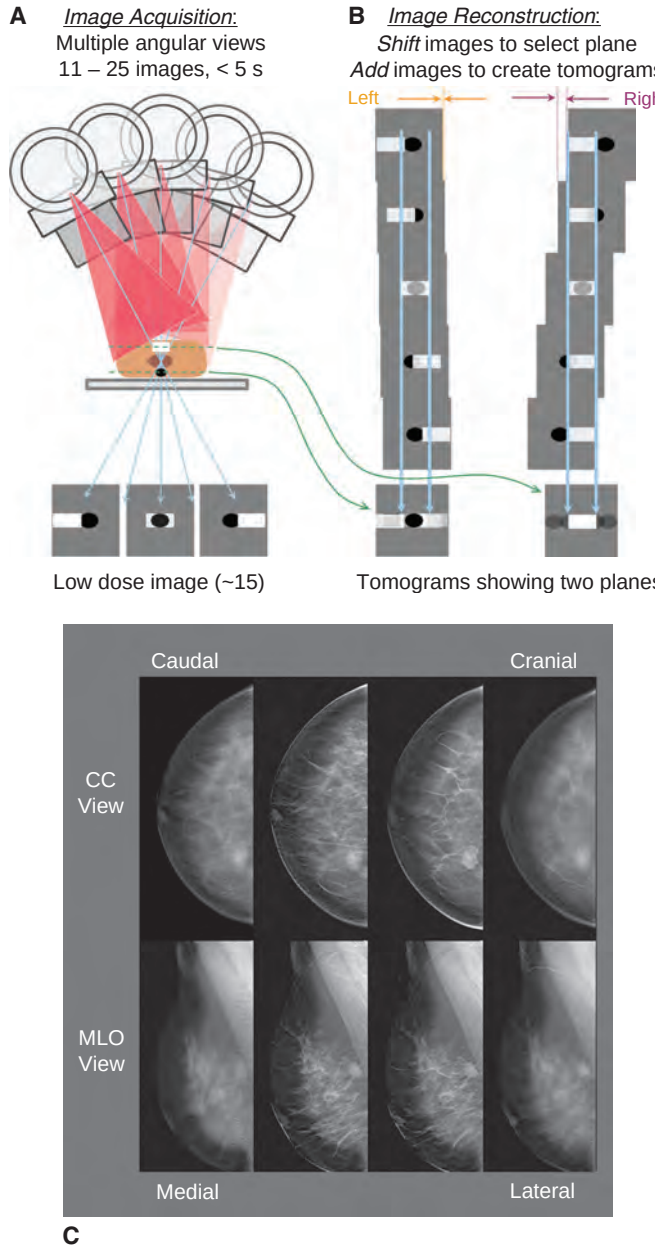
Digital add-on biopsy units with larger FOVs (e.g., 8 × 8 cm) using CCDs or other photosensitive detectors that fit into the cassette assemblies of the standard mammography units are alternatives to the prone biopsy table, as well as full-field digital mammography units that can provide similar capabilities.

Breast Digital Tomosynthesis

Digital radiographic imaging offers postacquisition processing capabilities that are not possible with conventional analog imaging systems. One of the major problems with conventional projection imaging is that overlying and underlying anatomy are superimposed on the pathology, often obscuring the visualization and detection of the cancer or other abnormality. A method for reducing this superimposition is to acquire multiple low-dose images at several angular positions as the x-ray tube moves in an arc about the breast. Each image projects the content in the breast volume with different shifts depending on the distance of the object from the detector. With high-speed digital readout, several projection images can be acquired in under 10 s, and each image set is processed with a limited-angle reconstruction algorithm to synthesize a tomogram (in-focus plane) at a given depth in the breast. Many tomograms representing incremental depths within the volume are produced, and in-focus planes throughout the breast can assist in enhanced detection of pathology (e.g., tumor) or elimination of superimposed anatomy that mimics pathology. Even though many more images must be reviewed by the radiologist, digital tomosynthesis technology can lead to a superior diagnosis that might save a patient an unneeded biopsy or provide guidance for early treatment of a cancer.

Digital tomosynthesis for breast imaging is made possible with fast readout TFT arrays by acquiring many sequential images (11 to 51) over a limited angle (± 7.5 to ± 50 degrees) in a short time span (4 to 20 s). For one vendor's implementation, 15 projection images of the compressed breast are acquired over a range of 15 degrees (1 image per degree) in 4s (270 ms per image), using a W target with 0.7-mm Al filtered x-ray beam, 32 to 38 kV, with no antiscatter grid, and 2 × 2 pixel binning of the digital detector array (140-μm sampling). The image acquisition process is illustrated in Figure 8-30A. Average glandular dose (discussed below) for 15 image acquisitions is approximately 2 mGy for the accreditation phantom. Larger breast thicknesses will require larger doses for tomosynthesis acquisition, but the dose is similar to that of a single view mammogram of the same compressed thickness. The 15 images are used in the reconstruction process to create 1-mm focal plane images (tomograms), where the location of the reconstructed plane depends on the reconstruction algorithm, and the number of reconstructed images depends on the compressed breast thickness. Reconstruction can be performed in several ways: (a) "shift and add" method shown in Figure 8-30B is the simplest; (b) filtered backprojection; (c) simultaneous algebraic reconstruction techniques; and (d) maximum likelihood algorithms. More sophisticated methods require longer reconstruction times but provide enhancement by further reducing out-of-plane signals.

A series of approximately 60 tomograms was reconstructed from a volunteer patient who was imaged in the cranial-caudal and medial-lateral oblique projections; four images from that dataset, at the entrance and exit points of the x-ray beam and at two depths, are shown in Figure 8.30C. Rapid "stack mode" viewing helps the perception of otherwise undetected lesions with good conspicuity. Investigations of breast tomosynthesis have demonstrated that its advantages are greater for masses and architectural distortion, than for calcifications. The current role of tomosynthesis is to supplement full-field digital mammography exams; however, future limited-angle acquisition techniques and processing algorithms are expected to provide both the tomosynthesis images at in-focal planes throughout the breast and a high-quality reconstructed conventional projection mammogram from the same image dataset.



■ **FIGURE 8-30** **A**. Digital tomosynthesis images are acquired using low-dose techniques over a limited projection angle, from ±7.5 degrees up to ± 30 degrees, with 15 to 40 images comprising the dataset. Illustrated are a "square" and "circle" at two depths within the breast **B**. Reconstruction of tomogram images at 1-mm incremental planes in the breast is performed by shifting the acquired images a known amount, summing the images by projection and normalizing the output. In-focus images are shown for two depths, illustrating the ability to blur the underlying and overlying signals for the white rectangle and black circle. **C**. Tomosynthesis images from a volunteer patient are shown for the cranial-caudal and medial-lateral oblique projections at 4 depths. The overall breast dose is similar to that from a single projection mammogram.

8.6 Radiation Dosimetry

X-ray mammography is the technique of choice for detecting nonpalpable breast cancers. However, the risk of carcinogenesis from the radiation dose to the breast is of concern, particularly in screening, because of the very large number of women receiving the exams. Thus, the optimization of breast dose is important and dose monitoring is required yearly as specified by the MQSA regulations.

The glandular tissue is the site of carcinogenesis, and thus the preferred dose index is the *average glandular dose*. Because the glandular tissues receive varying doses depending on their depths from the skin entrance site of the x-ray beam, estimating the dose is not trivial. The *midbreast dose,* the dose delivered to the plane of tissue in the middle of the breast, was the radiation dosimetry benchmark until the late 1970s. The midbreast dose is typically lower than the average glandular dose and does not account for variation in breast tissue composition.

Average Glandular Dose

The average glandular dose, D_g, is calculated from the equation

$$D_g = D_{gN} \times X_{ESAK}$$

where X_{ESAK} is the entrance skin air kerma (ESAK) in mGy, and D_{gN} is an air kerma to average glandular dose conversion factor with units of mGy dose/mGy incident air kerma. An air ionization chamber is used to measure the air kerma for a given kV, mAs, and beam quality. Previously, this calculation involved a roentgen to rad conversion, where the units of the D_{gN} were mrad/R.

The conversion factor D_{gN} is determined by computer simulation methods and depends on radiation quality (kV and HVL), x-ray tube target material, filter material, breast thickness, and tissue composition. Table 8-7 lists D_{gN} values for a 50% adipose, 50% glandular breast that is 4.5-cm thick. The coefficients are shown as a function of HVL and kV for a Mo/Mo target-filter combination. For a 26-kV technique with a HVL of 0.35-mm Al, the average glandular dose is approximately 19% of the measured ESAK (Table 8-7). For higher average x-ray energies (e.g., Mo/Rh, Rh/Rh, W/Rh, W/Ag), conversion tables specific to the generated x-ray spectrum must be used, as the D_{gN} values increase due to the higher effective energy of the beam. D_{gN} values decrease with an increase in breast thickness for constant beam quality and breast composition. This is because the glandular tissues furthest from the beam entrance receive much less dose in the thicker breast due to attenuation (e.g., D_{gN} = 0.220 mGy dose/mGy air kerma for 3-cm thickness versus 0.110 mGy dose/mGy air Kerma for 6-cm thickness). However, the lower D_{gN} coefficients for the thicker breast do not mean that larger breasts receive less dose. The lower D_{gN} values are more than offset by a large increase in the entrance kerma necessary to achieve the desired OD on the film or SNR in a digital image. Measured average glandular doses for 50% glandular/50% adipose tissue of 2 to 8-cm tissue thicknesses using AEC are shown in Figure 8-31 for screen-film and direct-detection digital systems.

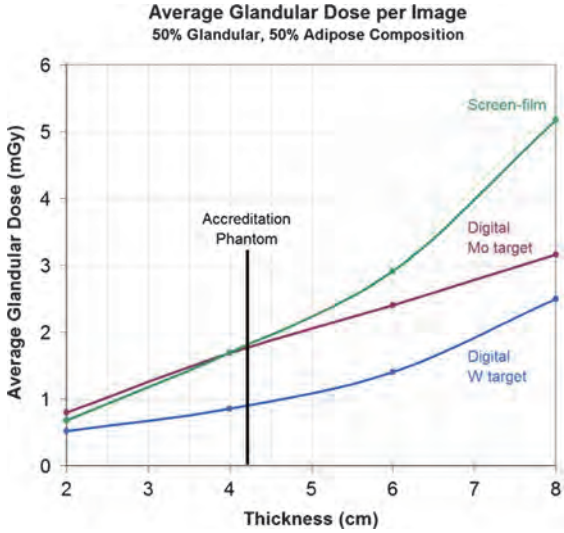
Factors Affecting Breast Dose

Many variables affect breast dose. The speed of the screen-film receptor, the film OD, and the digital detector SNR level are major factors. Breast thickness and tissue composition strongly affect x-ray absorption. Higher kV (higher HVL)

TABLE 8-7 D_{gN} CONVERSION FACTOR (mGy AVERAGE GLANDULAR DOSE PER mGy INCIDENT AIR KERMA) AS A FUNCTION OF HVL AND kV FOR Mo TARGET AND FILTER, 4.5-cm BREAST THICKNESS OF 50% GLANDULAR AND 50% ADIPOSE BREAST TISSUE COMPOSITION

	kV							
HVL (mm)	25	26	27	28	29	30	31	32
0.25	0.140							
0.26	0.144	0.147						
0.27	0.149	0.151	0.153					
0.28	0.153	0.156	0.158	0.159				
0.29	0.159	0.161	0.163	0.164	0.165			
0.30	0.164	0.166	0.167	0.168	0.169	0.171		
0.31	0.168	0.171	0.172	0.173	0.174	0.175	0.176	
0.32	0.173	0.175	0.176	0.177	0.179	0.181	0.182	0.183
0.33	0.177	0.180	0.181	0.182	0.183	0.185	0.187	0.188
0.34	0.183	0.184	0.185	0.187	0.188	0.190	0.191	0.192
0.35	0.188	0.190	0.191	0.192	0.194	0.195	0.196	0.197
0.36	0.192	0.195	0.196	0.197	0.198	0.199	0.200	0.202
0.37		0.199	0.200	0.202	0.203	0.204	0.204	0.205
0.38			0.205	0.206	0.207	0.208	0.208	0.210
0.39				0.211	0.212	0.213	0.213	0.214
0.40					0.216	0.218	0.219	0.220

Adapted from ACR QC Manual, 1999; converted from mrad per R using 0.01 mGy/mrad and 0.1145 mGy air Kerma/R



■ **FIGURE 8-31** Approximate average glandular doses (25 to 35 kV using AEC acquisition) are shown for a screen-film detector with Mo target and Mo/Rh filtration (green), a digital system using a-Se direct conversion detector with Mo target and Mo/Rh filtration (purple), and a digital system using a-Se direct conversion detector with W target and Rh/Ag filtration (blue). The accreditation phantom (4.5 cm thickness) is indicated with the vertical black bar for comparison of average glandular dose noted in MQSA reports.

TABLE 8-8 AVERAGE GLANDULAR DOSE IN mGy WITH TWO TISSUE COMPOSITIONS OF THE BREAST USING kV VALUES IN TABLE 8-4[a] FOR SCREEN-FILM AND TABLE 8-5[b] FOR DIGITAL ACQUISITIONS

	BREAST THICKNESS (cm)			
DETECTOR: BREAST COMPOSITION	2	4	6	8
Screen-Film: 100% Adipose	0.50	1.25	3.00	4.30
Screen-Film: 50/50	0.65	1.80	3.80	5.20
Digital Selenium, W target: 100% Adipose	0.50	0.75	1.55	2.45
Digital Selenium, W target: 50/50	0.60	1.00	2.40	4.20

[a]Mo/Mo target/filter, 5:1 grid, and film OD = 1.80 (Fuji AD-M film and AD-fine screens).
[b]W/Rh (2, 4, 6 cm); W/Ag (8 cm) for selenium detector.

increases beam penetrability (lower ESE and therefore lower average glandular dose) but also decreases subject contrast. Antiscatter grids improve subject and radiographic contrast, but increase radiation dose by the Bucky or digital technique compensation factor (~2×). Table 8-8 lists measured average glandular doses for a contemporary clinical screen-film mammography system with a film OD of 1.8, a Mo/Mo target/filter, and a kV recommended in Table 8-4, compared to a digital mammography system utilizing a W/Rh or W/Ag target/filter combination with kV recommended in Table 8-5. Digital acquisition with higher effective energy beams is shown to substantially reduce average glandular dose over that of screen-film systems.

The MQSA regulations limit the average glandular dose for a compressed breast thickness of 4.2 cm and a breast composition of 50% glandular and 50% adipose tissue (the MQSA-approved mammography phantom) to 3 mGy per film or digital image (6 mGy for two films or images). If the average glandular dose exceeds 3 mGy for this size breast, mammography may not be performed. The average glandular dose using screen-film receptors is typically about 2 mGy per view or 4 mGy for two views. For full-field digital mammography systems, the dose is lower, typically from 1.2 to 1.8 mGy per view.

Mammography image quality strongly depends on beam energy, quantum mottle, and detected x-ray scatter. Faster screen-film detectors can use a lower dose, but often at the expense of spatial resolution. Digital detectors with higher quantum detection efficiency can deliver significantly reduced dose; the tradeoff for reduced dose can be a loss of SNR and contrast-to-noise ratio (CNR) and potentially less diagnostic information. Higher energy beams (e.g., Rh target Rh filter or W target Rh filter or Ag filter) can reduce the dose to thick or dense glandular breasts with little or no loss of image quality.

8.7 Regulatory Requirements

Regulations mandated by the MQSA of 1992 specify operational and technical requirements necessary to perform mammography in the United States. These regulations are contained in Title 21 of the Code of Federal Regulations, Part 900.

Accreditation and Certification

Accreditation and certification are two separate processes. For a facility to perform mammography legally under the MQSA, it must be accredited and certified. To begin the process, the facility must first apply for accreditation from an accreditation body (the ACR or one of several states if located in one of those states). The accreditation body verifies that the mammography facility meets standards set forth by the MQSA to ensure safe, reliable, high-quality, and accurate mammography. These include the initial qualifications and continuing experience and education of interpreting physicians, technologists, and physicists involved in the mammography program. Existence of an active QC program and image quality standards are verified and validated. *Certification* is the approval of a facility by the US FDA to provide mammography services and is granted when accreditation is achieved.

Specific Technical Requirements and Quality Control

For screen-film mammography systems, requirements are largely unchanged from the 1999 FDA regulations. Tables 8-9 and 8-10 summarize the annual QC tests performed by the technologist and physicist, respectively, as required by MQSA regulations.

Under FDA rules, facilities using x-ray systems other than screen-film mammography must maintain a QA and QC program that is substantially equivalent to the program recommended by the manufacturer. This has created a number of unique tests that the technologist and physicist must understand and perform according to the manufacturer's manuals and procedural methods. In addition, QC procedures for the diagnostic review workstation monitor and the laser printer for producing hard-copy film are also under the auspices of the manufacturer.

Federal regulations specify that whenever QC test results fall outside of the action limits *for image receptor modalities other than screen-film*, the source of the problem shall be identified and corrective action taken before any further examinations are performed. However, an alternative QC standard lists three different categories in which specific action requirements are detailed.

Category A QC tests and action levels are specific to the digital acquisition system and, when outside of the prescribed limits, the source of the problem must be identified and corrective action taken before any examinations can be performed. These tests include system resolution, breast dose, image quality evaluation, SNR/CNR, flat-field calibration, and compression.

Category B QC tests and action levels are specific to the performance of a diagnostic device used for mammographic interpretation, including the review workstation and laser film printer. When tests produce results that fall outside of the action limits specified by the manufacturer, the source of the problem shall be identified and corrective action taken before that specific device can be used for interpretation. Image acquisition can continue, as long as there are alternate approved diagnostic devices available for interpretation.

Category C QC tests and action levels evaluate performance of components other than the digital image receptor or diagnostic devices for interpretation. When test results are outside the action limits specified by the manufacturer, the source of the problem shall be identified and corrective action shall be taken within 30 days of the test date. Clinical imaging and mammographic image interpretation can continue during this period. Applicable QC tests include collimation assessment, artifact evaluation,

TABLE 8-9 PERIODIC TESTS PERFORMED BY THE QC TECHNOLOGIST FOR SCREEN-FILM DETECTOR SYSTEMS

TEST AND FREQUENCY	REQUIREMENTS FOR ACCEPTABLE OPERATION	DOCUMENTATION GUIDANCE	TIMING OF CORRECTIVE ACTION
Daily: Processor	Established operating level	QC records 12 mo QC films 30 d	Before any clinical film processing is performed
Base + fog density	≤ +0.03 OD		
Middensity	±0.15 OD		
Density difference	±0.15 OD		
Weekly: Phantom	Established operating level	QC records 12 mo Phantom images 12 wk	Before any further exams are performed
Center OD	OD ≥ 1.20		
Reproducibility	±0.20 OD		
Density difference between disk and background	±0.05 OD		
Phantom scoring	Four fibers, three speck groups, three masses		
Quarterly:			
Fixer retention; quarterly	Residual fixer no >5μg/cm²	QC records since last inspection	30 d
Repeat analysis; quarterly	<2% change up or down from previous		
Semiannually:			
Darkroom fog	OD difference ≤ 0.05	QC records since last inspection	Before any clinical film processing
Screen-film contact	40 mesh copper screen with no obvious artifacts	QC records since last inspection, three previous contact tests	Before any further exams performed using cassettes
Compression device	Compression force ≥111 newtons (25 lb)	QC records since last inspection or past three tests	Before exams performed using compression device

kV accuracy, HVL assessment, AEC performance and reproducibility, radiation output rate, and compression thickness indicator.

An effort to standardize QC in digital mammography is underway, under the auspices of the ACR, based upon review of QC test results and findings from the multi-institutional Digital Mammography Imaging Screening Trial completed in 2005.

The Mammography Accreditation Phantom

The mammography accreditation phantom simulates the radiographic attenuation of an average-size compressed breast and contains structures that model very basic image characteristics of breast parenchyma and cancer. Its role is to help determine the adequacy of the overall imaging system (including film processing and digital processing) in terms of detection of subtle radiographic findings, and to assess the reproducibly of image characteristics (e.g., contrast and OD or SNR and CNR) over time. It is composed of a poly-methyl-methacrylate (PMMA) block, a wax insert, and a PMMA disk (4-mm thick, 10-mm diameter) attached to the top of the phantom, and is intended to mimic the attenuation characteristics of a "standard breast" of 4.2-cm compressed breast thickness of 50% adipose and 50% glandular tissue composition. The wax insert contains six

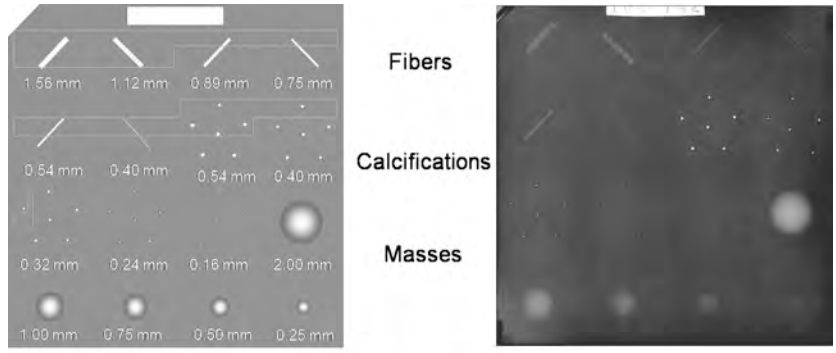
TABLE 8-10 SUMMARY TABLE OF ANNUAL QC TESTS BY THE MEDICAL PHYSICIST FOR SCREEN-FILM DETECTOR SYSTEMS

TEST	REGULATORY ACTION LEVELS	TIMING OF CORRECTIVE ACTION
AEC	OD exceeds the mean by >0.30 (over 2–6 cm range) or the phantom image density at the center is <1.20	30 d
kV	>5% of indicated or selected kV; C.O.V. > 0.02	30 d
Focal spot	See Table 8.1	30 d
HVL	See Tables 8.2 and 8.3	30 d
Air kerma and AEC reproducibility	Reproducibility C.O.V.[a] > 0.05	30 d
Dose	>3.0 mGy per exposure	Immediate, before any further exams
X-ray field/light field congruence	>2% SID at chest wall	30 d
Compression device alignment	Paddle visible on image	30 d
Screen speed uniformity	O.D. variation >0.30 from the maximum to minimum	30 d
System artifacts	Determined by physicist	30 d
Radiation output	<6.1 mGy/s (700 mR/s)	30 d
Automatic decompression control	Failure of override or manual release	30 d
Any applicable annual new modality tests	To be determined	Immediate, before any further exams

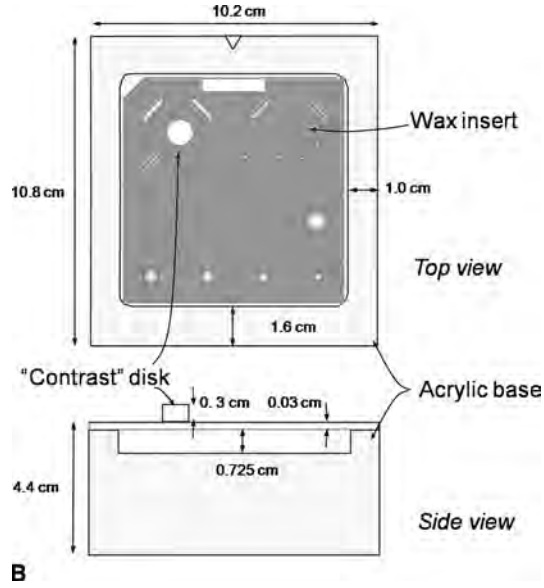
Note: documentation requires the inclusion of two most recent reports
[a]C.O.V., coefficient of variation, equal to the standard deviation of a series of measurements divided by the mean.
AEC, automatic exposure control; SID, source-to-image distance.

cylindrical nylon fibers of decreasing diameter, five simulated calcification groups with simulated calcifications (Al_2O_3 specks) of decreasing size, and five low contrast disks of decreasing diameter and thickness that simulate masses (Fig. 8-32A). Identification of the smallest objects of each type that are visible in the phantom image indicates system performance. To pass the MQSA image quality standards for screen-film mammography, at least four fibers, three calcification groups, and three masses must be clearly visible (with no obvious artifacts) at an average glandular dose of less than 3 mGy. Further details on other parameters can be found in the MQSA regulations.

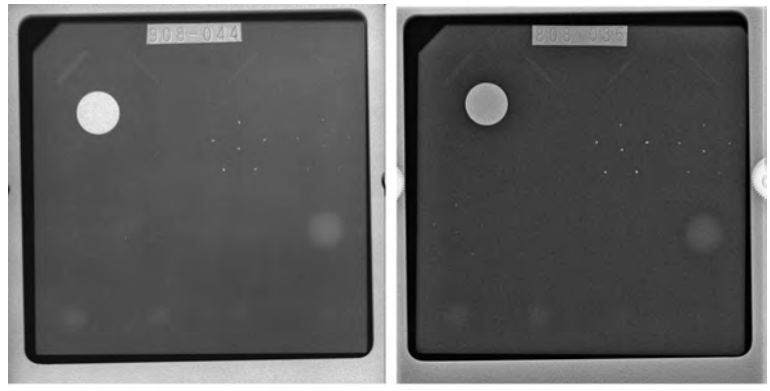
For digital mammography, each manufacturer has set the criteria for meeting an acceptable performance level in terms of qualitative visibility of the objects in the accreditation phantom, as well as specific tests using the quantitative evaluation of regions of interest (ROI) to calculate SNR and CNR. The "direct flat panel" detector manufacturer, for instance, requires the visibility of five fibers, four speck groups, and four masses, whereas the "indirect flat panel" and the cassette-based CR detector manufacturers require the visibility of four fibers, three speck groups, and three masses. Quantitative evaluation for one manufacturer includes the evaluation of the signal under the added acrylic disk and in the background area using ROIs of a specified size



A



B



C Screen-film Digital

■ **FIGURE 8-32** **A**. The mammography accreditation phantom contains a wax insert which has six nylon fibers, five aluminum oxide speck groups (six specks in each group), and five disks simulating masses. Mean diameters of the objects are indicated. The corresponding radiograph of the wax insert shows their "undegraded" radiographic appearance. **B**. The wax insert is placed into a cutout in a clear acrylic phantom with dimensions listed in the diagram. The phantom is intended to mimic the attenuation characteristics of a "standard breast" of 4.2-cm compressed breast thickness of 50% adipose and 50% glandular tissue composition. Note the "contrast disk" on top of the phantom, which generates a signal for evaluation of contrast. **C**. Mammographic image of the composite phantom with scatter and noise illustrates the loss of image quality.

and locations, while another uses the largest mass in the ACR phantom and an adjacent background area. The SNR is calculated as the average signal in the background ROI divided by its standard deviation. The CNR is determined as the difference of the average signals from the object and background, divided by the standard deviation in the background ROI. Each manufacturer requires the mammography site technologist to measure the SNR and CNR weekly using the accreditation phantom, and to verify that the calculated values are within the manufacturer's established limits. When found out of tolerance, the system may not be used clinically until the causes of the failures are determined and the corrections are implemented and verified with subsequent tests.

Quality Assurance

Mammography is a very demanding imaging procedure, in terms of equipment performance, and in regards to technical competence of the radiologist, technologist, and medical physicist. Achieving the full potential of the mammography examination requires careful optimization of the technique and equipment. Even a small change in equipment performance, film processing, digital image SNR or CNR, image artifacts, patient setup, or image-viewing conditions can decrease the sensitivity and specificity of mammography.

Ultimate responsibility for mammography quality assurance rests with the radiologist in charge of the mammography practice, who must ensure that all interpreting radiologists, mammography technologists, and the medical physicist meet the initial qualifications and maintain the continuing education and experience required by the MQSA regulations. Records must be maintained of employee qualifications, quality assurance, and the physicist's tests. Clinical performance and QC issues should be periodically reviewed with the entire mammography staff.

The technologist performs the examinations, that is, patient positioning, compression, image acquisition, and film processing and/or evaluation of digital image quality. The technologist also performs the daily, weekly, monthly, and semiannual QC tests. The medical physicist is responsible for equipment performance measurements before first clinical use, annually thereafter, and following major repairs, and for oversight of the QC program performed by the technologist.

SUGGESTED READING

ACR Mammography Quality Control Manual, 1999. American College of Radiology, Reston, VA.

ACR Practice Guideline for the Performance of Screening and Diagnostic Mammography. 2008. American College of Radiology, Reston, VA. Document is available at: http://www.acr.org/SecondaryMainMenuCategories/quality_safety/guidelines/breast/Screening_Diagnostic.aspx

Food and Drug Administration Mammography Program (MQSA). The MQSA regulations and other documentation are available on the Web at http://www.fda.gov/cdrh/mammography

Integrating the Healthcare Enterprise (IHE) Radiology Technical Framework. Mammography image (MAMMO) profile. Document is available at: http://www.ihe.net/Technical_Framework/upload/IHE_Rad1344TF_Suppl_MAMMO_T1_2006-04-13.pdf

Pisano E, Gatsonis C, Hendrick E, et.al. Performance of digital versus film mammography for breast-cancer screening. *N Engl J Med* 2005;353:1773–1783.

Practice Guideline for Determinants of Image Quality in Digital Mammography. 2007. American College of Radiology, Reston, VA. Document is available at: http://www.acr.org/SecondaryMainMenuCategories/quality_safety/guidelines/breast/image_quality_digital_mammo.aspx