# MAMMOGRAPHY Chapter 14

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## INTRODUCTION

Breast cancer is a major killer of women. The International Agency for Research on Cancer estimates that over 1.38 million women were diagnosed with breast cancer internationally in 2008, with over 458 000 deaths. The causes are not currently known; however, it has been demonstrated that mortality can be significantly reduced if disease is detected at an early stage.

## RADIOLOGICAL REQUIREMENTS FOR MAMMOGRAPHY

Mammography is a radiographic procedure optimized for examination of the breast. For many women, mammography is a highly effective means of detecting early-stage breast cancer. It is used both for investigating symptomatic patients (diagnostic mammography) and for the screening of asymptomatic women in selected age groups. A typical mammographic screening examination consists of one or two views of each breast. Common views include the cranial-caudal and mediolateral oblique, an example of which is shown in Fig. 9.1 (see also the Appendix to this book). While it is used primarily for the detection and diagnosis of breast cancer, mammography also has value in the presurgical localization of suspicious regions and the guidance of biopsies.

Breast cancer is detected on the basis of four types of signs on the mammogram:

(i) The characteristic morphology of a tumor mass, which can include irregular margins and spiculations;

(ii) Certain presentations of mineral deposits, visualized as specks called microcalcifications;
(iii) Architectural distortion of normal tissue patterns caused by the disease;
(iv) Asymmetry between corresponding regions of the left and right breasts.

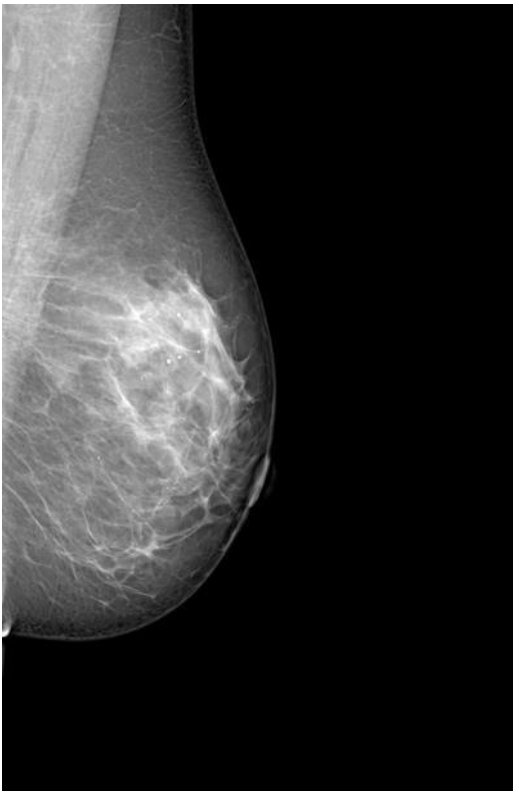

*FIG. 9.1. A mediolateral oblique mammogram. In this projection, the pectoralis muscle is visualized down to the level of the nipple. In this mammogram, characteristic benign calcifications are seen.*

Figure 9.2 shows X-ray attenuation coefficients measured versus energy on samples of three types of material found in the breast: adipose tissue, normal fibroglandular breast tissue and infiltrating ductal carcinoma (one type of breast tumor). Both the attenuation coefficients themselves, and their difference, decrease with increasing energy, resulting not only in a reduction in the radiation

dose required to produce an image, but also a decrease in image contrast. As shown in Fig. 9.3, the inherent X-ray subject contrast falls as X-ray energy increases. Note that the subject contrast of even small calcifications in the breast is similar to that for a tumor mass because of the greater difference in attenuation coefficient between calcium and breast tissue.

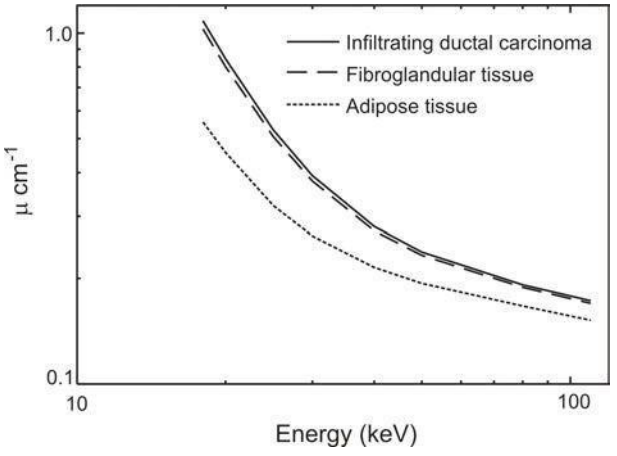


*FIG. 9.2. Dependence of the linear X-ray attenuation coefficient, μ, on X-ray energy.*

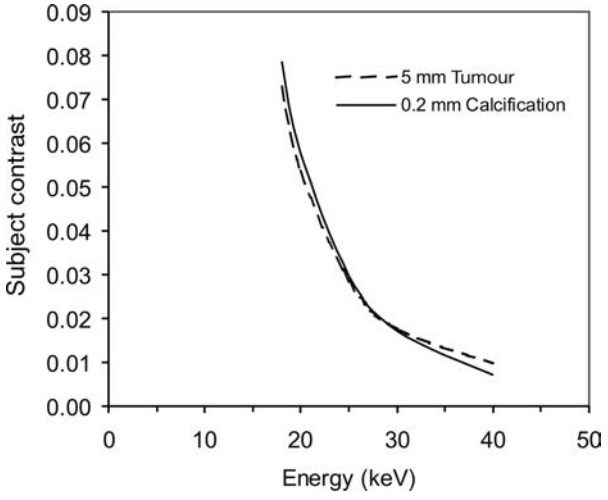


*FIG. 9.3. Dependence of image contrast on X-ray energy.*

The essential requirements for image quality in mammography represent a special case of the general principles of radiography outlined in Chapter 6, adapted to the specific imaging tasks involved in the detection of the radiological signs of breast cancer. The physics requirements are summarized here.

The imaging system must have sufficient spatial resolution at high spatial frequencies to delineate the edges of fine structures in the breast. Structural detail, possibly as fine as 50 μm, must be adequately resolved. Variation in X-ray attenuation among tissue structures in the breast gives rise to variation in the transmitted X-ray signal, and this is the fundamental source of image contrast.

As shown in Figs 9.2 and 9.3, breast tissues intrinsically lack subject contrast, requiring the use of low energy X-ray spectra, which emphasize the compositional differences of the breast tissues. Variation in the physical size and internal composition of the breast and age-dependent changes in the breast requires a broad dynamic range. The use of a radiographic grid and firm breast compression provides some compensation for scatter. The detectability of structures providing subtle contrast is further impaired by random fluctuations in the image, referred to as mottle or noise. However, the breast is sensitive to ionizing radiation, which, at least for high doses, has a small associated risk of inducing breast cancer. It is, therefore, desirable to use the lowest absorbed dose compatible with high diagnostic image quality. In the following sections, the specialized components of the mammographic imaging system will be described, and their design related to the above-mentioned imaging performance factors.

## X-RAY EQUIPMENT

The mammography unit consists of an X-ray tube and an image receptor mounted on opposite sides of a mechanical assembly. Because the breast must be imaged from different aspects, the assembly can be rotated about a horizontal axis, as shown in Fig. 9.4. To accommodate patients of different heights, the assembly elevation can be adjusted.

Unlike most general radiography equipment, which is designed such that the image field is centered below the X-ray source, in mammography, the system's geometry is arranged as in Fig. 9.5(a). Here, a vertical line from the focal spot of the X-ray source grazes the chest wall of the patient and intersects orthogonally with the edge of the image receptor closest to the patient. If the X-ray beam were centered over the breast as in Fig. 9.5(b), some of the tissue near the chest wall would not be imaged.

Radiation leaving the X-ray tube passes through a metallic spectral shaping filter, a beam-defining aperture and a plastic plate, which compresses the breast onto the breast support platform. Those X rays transmitted through the breast

and breast support are incident on a specially designed anti-scatter grid, and then are incident on the image receptor, where they interact and deposit most of their energy locally. In screen film and cassette-based digital mammography systems, a fraction of the X rays pass through the receptor without interaction and these X rays impinge upon the sensor of the automatic exposure control (AEC) mechanism of the mammography unit. In other digital mammography systems, the AEC mechanism is typically integral with the digital image receptor. In all systems, any remaining primary X rays are attenuated by a primary beam stop.

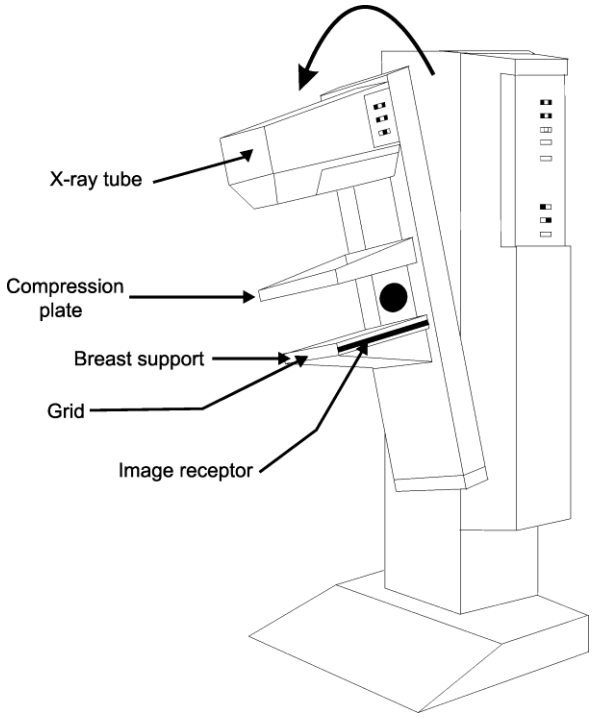


*FIG. 9.4. Schematic of a mammography imaging system.*

### Tubes, filters, and spectra

In modern mammography systems, the power supply is typical of the high-frequency type (see Section 5.4.2.3) and provides a nearly constant potential waveform during the exposure. The X-ray tube employs a rotating anode design in which electrons from the cathode strike the anode target material at a small angle from normal incidence (Fig. 9.6).

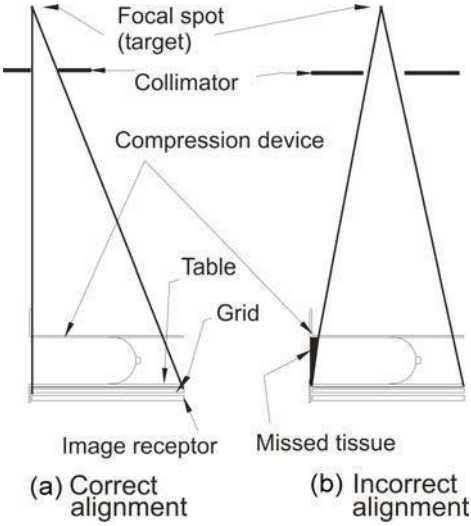


*FIG. 9.5. System geometry for image acquisition showing (a) correct alignment and (b) missed tissue associated with incorrect alignment.*

On modern equipment, the typical nominal focal spot size for contact mammography is 0.3 mm, while the smaller focal spot used primarily for magnification is 0.1 mm. The nominal focal spot size is defined relative to the effective spot size at a reference axis. As shown in Fig. 9.6, this reference axis, which may vary from manufacturer to manufacturer, is normally specified at some midpoint in the image. The effective size of the focal spot will monotonically increase from the anode side to the cathode side of the imaging field, as illustrated in Fig. 5.7. In mammography, the X-ray tube is arranged suchthat the cathode side of the tube is adjacent to the patient's chest wall, because thehighest intensity of X rays is available at the cathode side and the attenuation of X rays by the patient is generally greater near the chest wall. Frequently in breastimaging, there may be different target angles according to the focal spot size. In addition, the angulation of the X-ray tube itself may be changed according to thechoice of focal spot size and target material.

Most mammography tubes use beryllium exit windows between the evacuated tube and the atmosphere, and no oil is present in the radiation path exiting the tube. Oil, glass or other metals used in general purpose tubes would provide excessive attenuation of the useful energies for mammography.

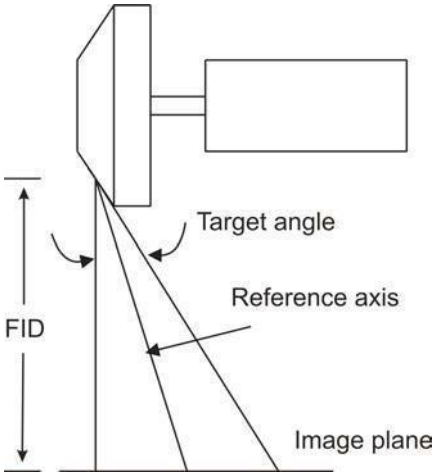


*FIG. 9.6. The geometry of an X-ray tube (FID: focus to image distance). The perpendicular line abuts the chest wall. The reference axis on a particular system will be specified by the manufacturer.*

As in general radiography, one attempts to define a spectrum that provides energies that give an appropriate compromise between radiation dose and image quality. In mammography, the spectral shape is controlled by adjustment of the tube voltage, choice of the target material and the type and thickness of the metallic filter placed between the X-ray tube and the breast. The strategies for optimization of the X-ray spectrum for screen-film mammography and digital mammography are quite different. In screen-film mammography, the contrast of the displayed image is constrained by the fixed gradient of the film, while in digital mammography, the quality of the displayed image is constrained by the image signal-to-noise ratio (SNR).

Using monoenergetic models of mammographic imaging, it has been suggested that the optimum energy for film imaging lies between 18 and 23 keV, depending on breast thickness and composition. It has been found that for a breast of typical thickness and composition, the characteristic X rays from molybdenum (see Fig. 1.3) and rhodium (see Table 9.1) provide good imaging performance for screen-film mammography. For this reason, molybdenum and/or rhodium target X-ray tubes are available on most mammography machines. Because the contrast of digital images can be controlled during image display, higher energiesmay be more optimal for digital mammography. For this reason, some digital mammography machines provide tubes equipped with tungsten targets.

TABLE 9.1. CHARACTERISTIC X-RAY ENERGY (keV) FOR MOLYBDENUM (Mo) AND RHODIUM (Rh) ANODE X-RAY TUBES

| Anode | $K_{\alpha 1}$ | $K_{\alpha 2}$ | $K_{\beta 1}$ |
|---|---|---|---|
| Mo | 17.48 | 17.37 | 19.61 |
| Rh | 20.22 | 20.07 | 22.72 |

As in conventional radiology, metallic filters are used in mammography to provide selective removal of low X-ray energies from the beam before it is incident upon the patient. In mammography, a molybdenum anode X-ray tubeis commonly employed with a molybdenum filter that is 30–35 μm thick. This filter acts as an energy window providing greater attenuation of X rays both at low energies and above the K absorption edge at 20 keV, while allowing the molybdenum characteristic X rays from the target and X rays of similar energy produced by bremsstrahlung to pass through the filter with relatively high efficiency. As illustrated by Fig. 9.7(a), the resultant spectra are enriched with X rays in the range of 17–20 keV.

Although molybdenum spectra are relatively well suited for imaging a breast of average attenuation, slightly higher energies are desirable for imaging thick, dense breasts. Because the molybdenum target spectrum is so heavily influenced by the characteristic X rays, an increase in the tube voltage alonedoes not substantially change the shape of the spectrum (see Fig. 9.7(a)). The average energy of the beam can be increased, however, by employing filters of higher atomic numbers than molybdenum. For example, rhodium (atomic number45) has a K absorption edge at 23 keV, providing strong attenuation both forX rays above this energy and for those at substantially lower energies. Usedwith a molybdenum target X-ray tube and slightly increased kV, it provides a spectrum with increased penetration (reduced dose) compared with the Mo/Mo combination. A Mo/Rh X-ray spectrum is illustrated in Fig. 9.7(b).

Further improvement in imaging performance can be obtained by tuning the effective spectral energy using other target materials in combination with appropriate K edge filters. One example is the use of an X-ray tube that incorporates a rhodium target. A 25–35 μm thick rhodium filter is used with this target material. Figure 9.7(c) illustrates the spectrum produced with an Rh target and an Rh filter. Similarly, particularly for digital mammography, K edge filtration of tungsten spectra can be used to advantage in that the lack of pronounced K characteristic peaks provides flexibility in spectral shaping with filters, as illustrated in Fig. 9.7(d). Typically, filters composed of aluminum, rhodium or silver are used to shape the tungsten spectrum.

For screen-film mammography, the fixed characteristic curve of the film imposes limitations on the suitable energy range; contrast and noise are limited

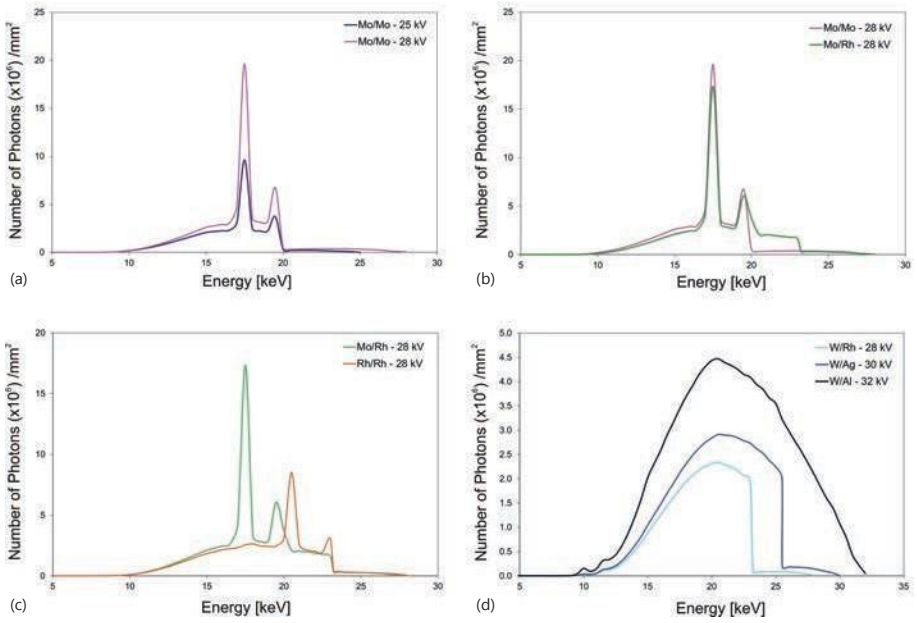


*FIG. 9.7. Examples of mammographic X-ray spectra.*

by the acceptable radiation dose to the breast. For digital mammography, the limitations imposed by the film are removed and the gradient of the image display is freely adjustable at the viewing workstation. This provides an opportunity to use higher energy beams to improve the SNR per unit dose and to provide the potential for dose reductions. For example, whereas a typical exposure technique for an average breast with screen-film mammography might be Mo target, Mo filter and 26 kV, with a digital mammography system, either a Mo/Rh or Rh/Rh combination might be used at 28 or 29 kV, or a tungsten target with Ag or Rh filtration operated at similar tube voltage.

### Compression

There are several reasons for applying firm (but not painful) compression to the breast during the mammographic examination. Compression causes the various breast tissues to be spread out, minimizing superposition from different planes and thereby improving the conspicuity of structures. This effect may be accentuated by the fact that different tissues (fatty, fibroglandular and cancerous) have different elasticities, resulting in the various tissues being spread out by different amounts and potentially making a cancer easier to see.

As in other areas of radiography, scattered radiation will degrade contrast in the mammogram. The use of compression decreases the ratio of scattered to directly transmitted radiation reaching the image receptor. In Fig. 9.8, the effect of breast thickness on scatter is quantified. Compression also decreases the distance from any plane within the breast to the image receptor, and in this way reduces geometric unsharpness. The compressed breast provides lower overall attenuation to the incident X-ray beam, allowing the radiation dose to be reduced. The compressed breast also provides more uniform attenuation over the image. This reduces the exposure range that must be recorded by the imaging system, and in screen-film mammography allows a film of higher gradient to be employed. Finally, compression provides a clamping action, which reduces anatomical motion during the exposure, thereby reducing this source of image unsharpness.

It is important that the breast be compressed as uniformly as possible and that the edge of the compression plate at the chest wall be straight and aligned with both the focal spot and image receptor to maximize the amount of breast tissue that is included in the image (see Fig. 9.5). The mechanical properties of the breast are non-linear; after a certain reduction in thickness, application of additional pressure provides little benefit in terms of improved image quality and only contributes to patient discomfort. Specialized mechanisms have been introduced by several manufacturers to try to achieve better compression, while minimizing the risk of over compression.

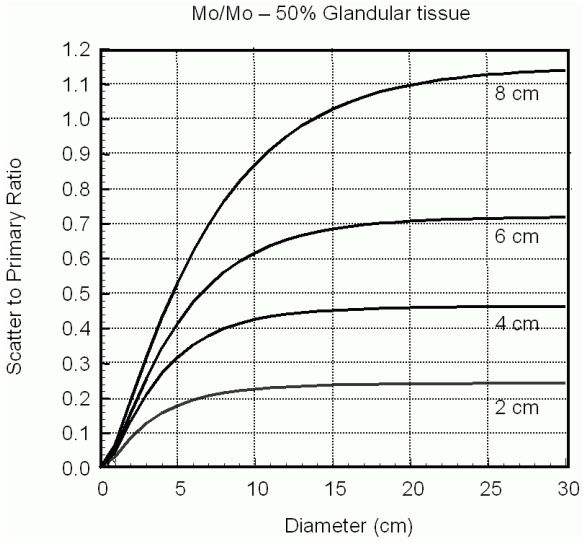


*FIG. 9.8. The effect of breast thickness and diameter of the X-ray field on the SPR when no anti-scatter grid is used.*

## Grids

In the absence of an anti-scatter device, 37–50% of the total radiation incident on the image receptor would have experienced a scattering interaction within the breast. Thus, as seen in Fig. 9.8, the scatter to primary ratio (SPR) will range from 0.3 to 1.2, depending upon the breast size. In addition to contrast reduction, the recording of scattered radiation reduces the useful dynamic range of the image receptor and adds stochastic noise to the image. The actual SPR recorded in the image is determined in part by the detector material; the lower energy and oblique incidence of scattered X rays result in higher attenuation than for primary X rays (see Chapter 6).

It is typical to use focused linear grids in mammography, with grid ratios from 3.5:1 to 5:1. On modern mammography equipment, the grid is an integral part of the system, and during the X-ray exposure is moved to blur the image of the grid septa to avoid distracting artifacts in the mammogram. It is important that this motion be uniform and of sufficient amplitude to avoid non-uniformitiesin the image, particularly for short exposures that occur when the breast is relatively radiolucent. At least one manufacturer provides a crossed grid that consists of septa that run in orthogonal directions. Improved scatter rejection is accomplished at doses comparable to those required with a linear grid, because the interspace material of the crossed grid is air rather than a solid. To avoid artifacts, the crossed grid must be moved in a very precise manner to ensure a uniform blurring.

When a grid is used, the SPR is typically reduced by a factor of about 5, leading in most cases to a substantial improvement in image contrast (Fig. 9.9(a)). As discussed in Chapter 6, to maintain image quality when the grid is used, it is necessary to compensate for losses of X-ray fluence at the image receptor that are caused by absorption of primary radiation by the grid septa and interspace material, as well as the removal of scatter by the grid. This is reflected in the Bucky factor, which can be as large as 2 to 3 in mammography (Fig. 9.9(b)).

The improvement in image contrast in screen-film mammography andSNR in digital mammography is generally considered to justify this increasein dose to the breast. Some differences do exist between digital and screen-film mammography. In mammography, the benefit of a grid is clear for thick breasts; however, in the digital mammography of small or thin breasts, the signaldifference to noise ratio (SDNR) improvement from scatter reduction may not justify the dose increase from the use of a grid. Also, for digital mammography,it is not necessary to compensate for the removal of scattered radiation. This allowsa reduction of the Bucky factor in digital imaging and a corresponding dose reduction.

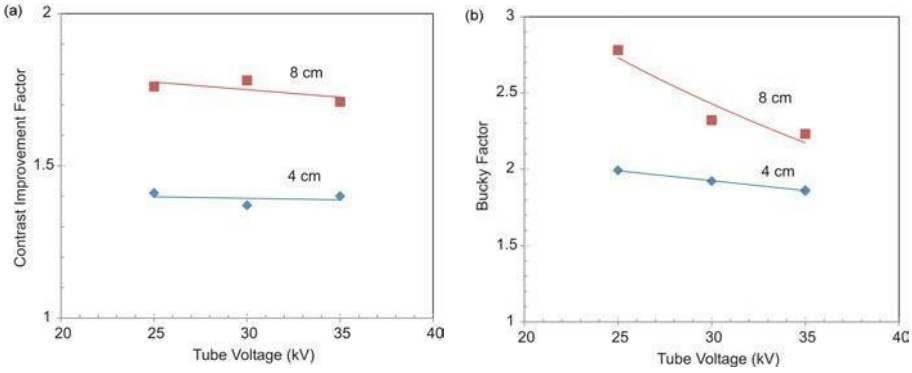


*FIG. 9.9. The effect of the use of a grid for screen-film mammography on (a) contrast improvement factor and (b) the Bucky factor. Curves are shown as a function of tube voltage for a Mo/Mo target filter combination with 4 cm and 8 cm thick breast equivalent phantoms.*

## AEC

It is difficult to estimate the attenuation of the breast by visual inspection; therefore, modern mammography units are equipped with AEC (see Sections 5.4.3, 6.2.7 and 19.5.2). For screen-film mammography, it is very important for both image brightness and contrast to achieve a target optical density (OD) in the image, while with digital mammography, it is more useful to achieve a target SNR or, preferably, a target SDNR in the image.

For screen-film mammography and for cassette-based digital systems, the AEC radiation sensor(s) is/are located behind the image receptor to avoid casting a shadow on the image. The sensors measure the X-ray fluence transmitted through both the breast and the image receptor and provide a signal to discontinue the exposure when a preset amount of radiation has been received by the image receptor. The location of the sensor is adjustable so that it can be placed behind the appropriate region of the breast to obtain proper exposure. The AEC performance must be independent of variations in breast attenuation, tube voltage or filter settings, and field size. With modern equipment, AEC is generally microprocessor-based, so that relatively sophisticated corrections can be made during the exposurefor the above effects and for reciprocity law failure of the film (see Section 7.3.5).Penetration through the breast depends on both breast thickness and composition. For a breast that is thick or dense, it is possible that for a relativelylow tube voltage, a very long exposure time would be required to achieveadequate film darkening or digital signal. This would result in a high dose tothe breast and possibly blur due to anatomical motion, while a more penetratingbeam allows a lower dose to be used but with a loss of image contrast. Thus,

many mammography AEC systems also incorporate automatic control of the tube voltage or target/filter/tube voltage combination. These systems sense the compressed breast thickness and the transmitted exposure rate and employ an algorithm to choose automatically the X-ray target and/or beam filter as wellas the tube voltage. Typically, a short (usually <100 ms) X-ray pre-exposure is madefirst. The recorded X-ray signal and the breast thickness measured fromthe compression plate position are used to infer the composition of the breast and determine the optimal exposure conditions.

To provide greater flexibility, multiple user-selectable algorithms can be incorporated into the system to weight the exposure factor selection towards either lower dose or higher image quality, according to the requirements ofthe examination. All systems build in appropriate constraints to ensure that the operation of the equipment is in compliance with applicable regulatory radiation dose limits and within the functional limitations of the X-ray tube, generator and image receptor. These requirements may cause the image quality to be less than optimal under certain circumstances.

For digital mammography, it is typical for the pre-exposure concept described above to be used, but in this case, rather than a single sensor measurement, an entire low dose image is created in the digital detector. This image can be analyzed to determine the overall SDNR or the minimum value over a set of small (~1 $cm^2$) regions of interest in the image. Then, the target material, the filter, and the tube voltage can be selected automatically to attempt to ensure a desired SDNR when the main exposure is performed.

As digital detectors can be operated at a range of input dose levels, it is possible to optimize imaging according to a priority of SDNR, lowest dose or a combination. Different manufacturers have approached this challenge in different ways and development in this area is ongoing. For example, the location of the edges of the breast can be determined automatically so that the algorithm is only sensitive to the region of the image within the breast. The algorithm canbe 'trained' to identify automatically the critical areas that will dominate the selection of exposure parameters. Special modes of operation can be developed for tasks such as imaging breasts containing implants. Again, these systems must be constrained by radiation regulations.

## Magnification mammography

Magnification mammography is often used intentionally to improve the diagnostic quality of the image. This is accomplished by elevating the breast above the image receptor, in effect reducing the focus to object distance and increasing the distance from the object to the image receptor. Magnification mammography

achieves three key benefits: (i) increased SNR, (ii) improved spatial resolution and (iii) dose efficient scatter rejection. These benefits are illustrated in Fig. 9.10.

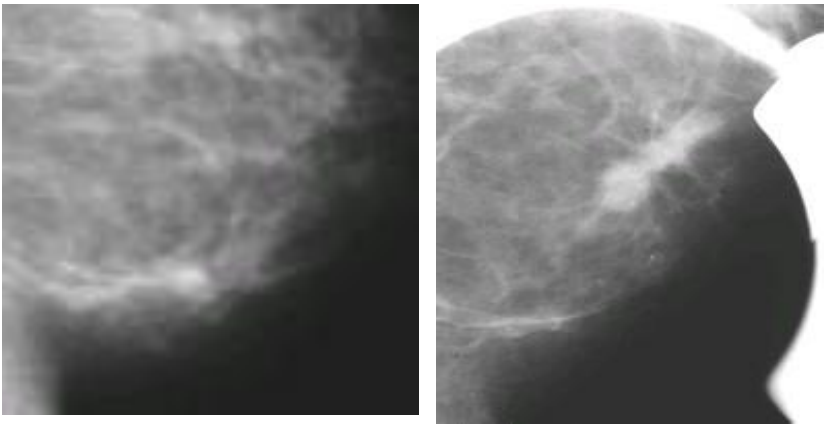

*FIG. 9.10. A suspicious region is visible in the lower aspect of the mammogram (left). A magnified image of this region obtained with focal compression shows an obvious mass (right).*

Magnification causes structures to appear larger when projected onto the image receptor, thereby increasing the effective modulation transfer function (MTF) of the receptor with respect to structures within the breast(see Sections 6.2.3 and 6.2.4). In screen-film mammography, the limiting resolution of the image receptor is already quite high and is rarely a limiting factor.The increase in focal spot unsharpness that occurs as a result of magnification, even with a small focal spot, typically offsets any improvement in the MTF of the image receptor. The main benefit of magnification is to increase the size of the projected anatomical structures compared with the granularity of the image, thereby improving the SNR in the image. This improvement can be valuable, particularly for the visualization of fine calcifications and spiculations.

In digital mammography, where the film grain noise has been eliminated, but where the limiting spatial resolution of the detector is lower than that provided by the screen film image receptor, the benefits of magnification may be different in nature. In this case, the increase in projected size of anatomical features does improve the effective resolution of the detector, which in some cases is a limiting factor.

The spatial resolution of magnification mammography is always limited by focal spot size. As such, the use of a small spot for magnification imaging (typically a nominal size of 0.1 mm) is critical. Loss of spatial resolution can be

controlled in part by using the minimum necessary magnification. It is typical to use magnifications of between 1.5 and 1.8. For the small focal spot, the X-ray tube current must be reduced, necessitating increased exposure times. As a result, there is an increased likelihood of resolution loss resulting from motion ofanatomical structures. It is common to apply focal compression to the breast in magnification imaging (Fig. 9.10, right), reducing the breast thickness and henceexposure time.

By moving the breast closer to the X-ray source in magnification mammography, the dose to breast tissue increases compared with that in contact mammography. The increased air gap between the breast and the image receptor provides some scatter rejection; thus, anti-scatter grids are not employed for magnification. This partially offsets the increase in dose and the increase in exposure time that occurs from use of the small focal spot.

## IMAGE RECEPTORS

### Screen film mammography

In screen-film mammography, a high-resolution fluorescent intensifying screen is used to absorb the X rays and convert the pattern of X rays transmitted by the breast into an optical image (see Section 7.3). These screens are used in conjunction with single emulsion radiographic film, enclosed within a lightproof cassette. The film is typically available in two sizes: 18 cm × 24 cm and24 cm × 30 cm. It is customary to use the smallest possible size that ensures complete radiographic coverage of the breast; this results in superior breast positioning and compression. In women with large breasts, multiple films may berequired to image the breast fully.

The screen and film are arranged as shown in Fig. 9.11, such that the X rays must pass through the cover of the cassette and the film to impinge upon the screen. Absorption is exponential, so that a larger fraction of the X rays is absorbed and converted to light near the entrance surface of the screen. The lateral spread of the light increases with the distance that light quanta travel through the phosphor. By minimizing the distance that the light must travel before being collected, this geometry reduces blurring due to lateral spreading, thus providing the maximal spatial resolution. To discriminate further against light quanta traveling along long oblique paths, the phosphor material of the screen may be treated with a dye, which absorbs much of this light, giving rise toa sharper image.

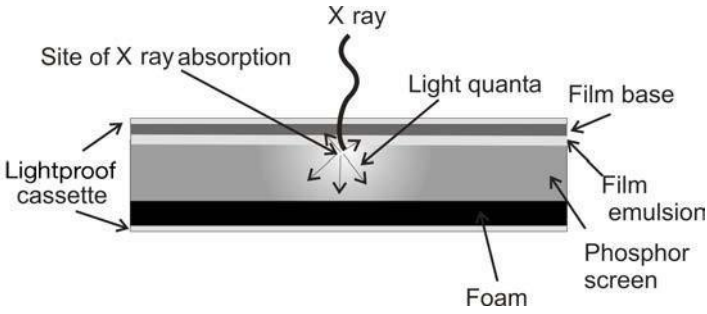


*FIG. 9.11. Configuration for a mammographic screen-film image receptor. A single emulsion radiographic film is held in close contact with a fluorescent screen in a lightproof cassette.*

A typical phosphor used for screen-film mammography is gadolinium oxysulphide ($Gd_2O_2S$:Tb). Although the K absorption edge of gadolinium occurs at too high an energy to be useful in mammography, the phosphor material is dense (7.44 g/cm$^3$), so the quantum detection efficiency (QDE) (the fractionof incident X rays that interacts with the screen (see Sections 7.2.1 and 7.3.2))is reasonably high (approximately 60% for a typical screen thickness and X-ray spectrum), and K fluorescence (a potential source of noise) is avoided. Also, the conversion efficiency (fraction of the absorbed X-ray energy converted to light) exceeds 10%, which is high for a phosphor. The amount of light emitted from the fluorescent screen is linearly dependent upon the total amount of energy deposited by X rays within the screen.

The photographic film emulsion for mammography is matched to be sensitive to the spectrum of light emitted from the particular phosphor screen, and to the range of X-ray fluence exiting the breast. As such, it is important to examine the overall characteristics of the screen and film combination rather thanthose of the individual components.

In mammography, compression of the breast reduces the overall range of X-ray fluence exiting the breast, as compared with a breast that is not uniformly compressed. This allows films with a high gradient to be used in an effort to enhance the contrast between subtly varying soft tissue structures (Fig. 9.12(a)). In addition, mammography film has a high maximum OD ($D_{max}$) (4.0–4.8 OD) to maximize the exposure latitude over which the high gradient exists (Fig. 9.12(b)). This is particularly important near the periphery of the breast where its thickness decreases rapidly. Nevertheless, some regions of the mammogram will generally be underexposed or overexposed, i.e. rendered with suboptimal contrast.

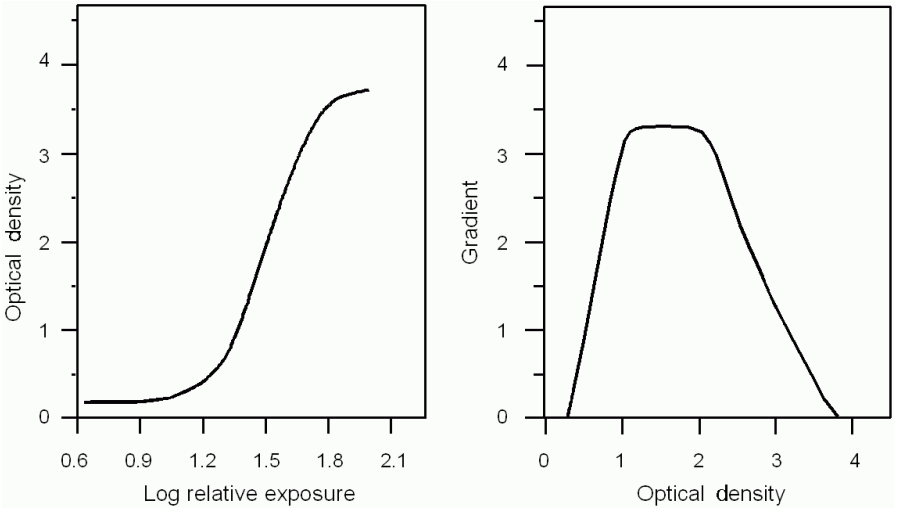


*FIG. 9.12. Characteristic curve of a film emulsion used for mammography.*

Mammography film is processed in an automatic processor similar tothat used for general radiographic films. It is important that the development temperature, time, and rate of replenishment of the developer chemistry be compatible with the type of film emulsion used and be designed to maintain good contrast of the film. Daily quality assurance is required in mammography toensure ongoing optimal performance.

There are several technical factors associated with screen-film mammography that limit the ability to display the finest or most subtle details and produce images with the most efficient use of radiation to the patient. In screen-film mammography, the film must act as an image acquisition detector aswell as a storage and display device. The sigmoidal shape of the characteristic curve results in limited latitude — the range of X-ray exposures over whichthe film display gradient is significant. If a tumor is located in a region of the breast that is either more lucent or more opaque, then the contrast displayed to the radiologist may be inadequate because of the limited gradient of the film. This is of particular concern in patients whose breasts contain large amounts of fibroglandular tissue, the so-called ‘dense breast’.

Another limitation of screen-film mammography is the effect of fixed pattern noise due to the granularity of the phosphor screen and the film emulsion used to record the image. This impairs the detectability of microcalcifications and other fine structures within the breast. Finally, screen-film mammography suffers from compromises in spatial resolution versus quantum detection efficiency, which are inherent in the screen film image receptor.

## Digital mammography

Digital mammography, introduced commercially in 2000, can overcome many of the technical limitations of screen-film mammography. In digital mammography, image acquisition, processing, display, and storage are performed independently, allowing optimization of each. Acquisition is performed with low noise X-ray detectors having a wide dynamic range. As the image is stored digitally, it can be displayed with contrast that is independent of the detector properties and defined by the requirements of the particular imaging task. Whatever image processing techniques are found useful, ranging fromsimple contrast enhancement to histogram modification and spatial frequency filtering, can conveniently be applied prior to image display.

The challenges in creating a digital mammography system with improved performance are mainly related to the X-ray detector and the display device. The detector should have the following characteristics:

- Efficient absorption of the incident radiation beam;
- A linear or logarithmic response over a wide range of incident radiation intensity;
- Low intrinsic noise and little to no fixed pattern noise, to ensure that images are X-ray quantum noise limited;
- Limiting spatial resolution of the order of 5–10 cycles/mm (50–100 μm sampling);
- It can accommodate at least an 18 cm × 24 cm and preferably a 24 cm × 30 cm field size;
- It can image immediately adjacent to the chest wall;
- An acceptable imaging time and heat loading of the X-ray tube (e.g. in detectors that must be scanned to image the entire breast).

Two main approaches have been taken in detector development — area detectors and scanning detectors. In the former, the entire image is acquired simultaneously, while in the latter only a portion of the image is acquired at one time and the full image is obtained by scanning the X-ray beam and detector(s) across the breast. Area detectors offer fast image acquisition and can be used with conventional X-ray machines equipped with a grid to reduce scatter. By comparison, scanning systems have longer acquisition times and are mechanically more complex, but use relatively simple detectors and have excellent intrinsic scatter rejection.

Various detector technologies are employed in full-field digital mammography systems, i.e. those capable of imaging the entire breast (see also Section 7.4.3). In one approach, the detector consists of an amorphous silicon

Thin-film transistor panel containing a rectangular matrix of 2000–3000 columns by 3000–4000 rows of detector elements (dels). Each del is connected by a thin film transistor switch to electrical lines running along each row and column (see Section 7.4.3). This array is covered by a phosphor or a photoconductor X-ray detector.

In so-called 'indirect' detectors, each del includes both a light-sensitive photodiode and a thin film transistor switch. The array is covered with a phosphor layer, typically made of thallium-activated CsI. X rays transmittedby the breast are absorbed by the phosphor and the light produced is converted in the photodiode to charge, which is stored on its capacitance. After the X-ray exposure, readout signals sent sequentially along the lines for each row activate the corresponding switches, and the charge is transferred down the columns to readout amplifiers and multiplexers, and digitized to form the image. This readoutsystem allows the signals from all of the dels to be read in a fraction of a second.The needle-like phosphor crystals of CsI (see also Fig. 8.2) behave somewhat like fiber optics, conducting the light to the photodiodes with less lateral spread than would occur with granular phosphors. This allows the thickness of the phosphor to be increased relative to a granular phosphor, to improve the QDE ofthe detector without excessive loss of spatial resolution.

A second system employs a similar readout strategy but replaces the phosphor with an X-ray absorber composed of amorphous selenium, which is a photoconductor. In this so-called 'direct' detector, the energy of the absorbed X rays causes the liberation of electron-hole pairs in the selenium. The charged particles are drawn to the opposite faces of the detector by an externally applied electric field. To collect the signal, an array of electrode pads (rather than photodiodes) forms the dels. Unlike the phosphor-based detectors, the electric field can be tailored to collect the charge with minimal lateral spread. This allows the use of a relatively thick detector to achieve excellent QDE without significant reduction in resolution at near-normal incidence. Other materials in which X-ray energy is directly converted to charge are under development and include lead iodide, zinc cadmium telluride, and thallium bromide. The use of higher atomic number materials would allow the thickness of the X-ray converter to be reduced. This mitigates against the degradation of the MTF resulting from the oblique incidence of the X rays.

Another technology used for digital mammography employs a plate formed of a photostimulable phosphor material, housed in a lightproof cassette, described in more detail in Section 7.4.2. When exposed to X rays, electrons in thecrystalline material are excited and subsequently captured by traps in the phosphor. The number of trapped electrons is proportional to the amount of X-ray energy absorbed at a particular location in the detector. After exposure,the plate is placed in a reader device and scanned with a red HeNe laser beam.

The energy of the laser light stimulates the traps to release the electrons. The transition of these electrons through energy levels in the phosphor crystal results in the formation of blue light. The light is collected by an efficient optical system, measured with a photomultiplier tube, and the signal digitized. By correlating the time of measurement of the signal with the position of the scanned laser beam, the signal can be attributed to a particular pixel in the image. The resolution of the image is determined by the size of the scanning laser beam, the underlying scatter of the readout laser light in the phosphor, and the distance between sample measurements.

Mammography photostimulable phosphor systems differ from the general radiography photostimulable phosphor systems in several key areas. In general, the mammography photostimulable phosphor system is designed for higher spatial resolution and thus uses a thinner phosphor material and is scanned with finer sampling pitch (typically 50 μm). The result is less signal per pixel. To overcome this limitation, various innovations have been developed to improve light coupling and reduce readout noise, including the use of dual-sided readout of the phosphor plates and needle-like phosphors that permit the use of thicker detectors with superior QDE.

The detector systems discussed thus far acquire the image by integrating the signal from a number of X-ray quanta absorbed in the detector and digitizing this signal. The image noise from these systems depends on both the PoissonX-ray quantum fluctuations associated with X-ray absorption and the additional noise sources associated with the production of the converted electronic signal. As discussed in Chapter 7, these noise sources can arise from the fluctuation in the amount of light produced in a phosphor in response to absorption of an X-rayof a particular energy, or from the X-ray spectrum itself (different amounts of signal are produced when X-ray quanta of different energies interact with the detector material).

As an alternative, it is also possible to count the number of interacting quanta directly, thereby avoiding these additional noise sources. Typically, quantum counting detectors are multiline devices employing a geometry in whichthe X-ray beam is collimated into a slot or multislit format and scanned across the breast to acquire the image. The detector can be based on either a solid-state approach, where electron-hole pairs are produced in a material such as crystallinesilicon, or a pressurized gas, where the signal is in the form of ions formed in thegas. In either case, collection of the charge signal and appropriate amplification producesa pulse for each interacting X-ray quantum and these pulses are simply counted to create the signal. An additional feature of these detectors is that asthe beam is collimated to irradiate only part of the breast at a time, the SPR is reduced without the need for a grid, and this increases the dose efficiency of the system.

## DISPLAY OF MAMMOGRAMS

### Display of film mammograms

To allow visualization of as much of the information recorded in the mammogram as possible, it is essential that the viewing conditions be optimal. Mammograms should be interpreted under conditions that provide good visibility and comfort and incur minimal fatigue. For film, viewing transillumination systems are available that have been specifically designed to produce the appropriate luminance levels for reading mammograms. The illuminator surface should provide diffused light of uniform brightness. The luminance level must be sufficient to illuminate areas of interest in the mammogram. It has been recommended that illuminators for film mammograms be capable of producing a luminance of at least 3000 $cd/m^2$.

The contrast sensitivity of the eye (the ability to distinguish small differences in luminance) is greatest when the surroundings are of about the same brightness as the area of interest. Therefore, to see detail in a mammogram, itis important to reduce glare to a minimum, to avoid surface reflections, and to reduce the ambient light level to approximately that reaching the eye throughthe mammogram. Glare and reflections can be reduced by locating illuminators away from bright surroundings such as windows, by switching off surrounding view boxes when not in use, and by using masks to cover unused portions of a view box or to cover low-density areas in the mammogram being examined.

Subdued lighting is preferred in the viewing room. It is also important to have a variable brightness high output light source (with appropriate masks) to view high OD areas on the film mammogram, and to ensure that films are properly exposed and processed.

### Display of digital mammograms

The display system plays a major role in influencing the overall performance of the digital mammography unit in terms of both the ease of image interpretation and the image quality presented to the radiologist. While some radiologistsuse ‘hard copy’ systems (laser printed films) for interpretation, the flexibilityof adjustment of display brightness and contrast in digital mammography isbest realized when the images are viewed on a computer display using either a cathode ray tube or flat panel display monitor. This is often referred to as a ‘soft copy’ display.

The display must have a suitable number of high-quality monitors (normally two 5 megapixel monitors are recommended) to allow viewing of as much of the mammogram as possible at the required resolution level. Digital

mammograms generally contain more pixels than can be displayed at one time on a soft copy device. A 5-megapixel monitor is capable of displaying only a single mammogram with approximately 2000 × 2500 100 μm pixels at full resolution. Larger images must be resampled to reduce temporarily the image size so that the entire mammogram can be viewed. Then, zooming and scrolling operations can be employed to allow inspection of regions of interest in the image at the full acquired spatial resolution.

Hard copy display systems produce a printout of the digital image on transparent film sensitized to laser light. The image brightness and contrast are usually adjusted by the radiographer before printing out the image, making use of the controls provided at the acquisition workstation. Hard copy image displays have the disadvantage of not allowing the radiologist to control the image processing operations during viewing. Therefore, it is strongly recommended that images be displayed for interpretation on a high-quality soft copy device.

Both wet and dry laser printers are available and produce breast images of similar quality. The spatial sampling (resolution) of laser printers should at least match the del size, so the printing device should not be the limiting factor. Using too low a resolution for printing results in printed films with coarse-looking pixels, or in an overly magnified image of the breast. Most of the commercially available printers for digital mammography have two-pixel sizes, 100 μm and50 μm or smaller (around 600 dpi), with a pixel bit depth of 12 or 14 bits, andcan print on one or more sizes of transparent film (typically 18 cm × 24 cm or24 cm × 30 cm).

Typically, transparent laser films do not have the same maximum OD capability as mammography films. The dynamic range of laser films varies from approximately 0.2 OD up to around 3.2 OD, depending on the film type. It is recommended that the laser printer characteristic curve be in conformance to the Digital Imaging and Communications in Medicine (DICOM) Grayscale Standard Display Function (GSDF) (see Chapter 16).

### *Image processing of digital mammograms*

In addition to resampling, zoom, and scrolling operations, several other types of image processing can be employed to improve the presentation of information in digital mammograms. Common image processing steps include: (i) segmentation of the breast from the background (air), so that the greyscale lookup table inversion does not display the air as white, (ii) peripheral enhancement to suppress the effect of changing thickness near the edges of the breast, (iii) image resolution restoration or enhancement to increase the conspicuity of the clinical signs of breast cancer, (iv) lookup table manipulation to allow adjustment of displayed brightness and contrast and to match the performance of the human

eye, and (v) noise suppression to enhance the relative conspicuity of small calcific deposits found in the breast. Methods of image processing are discussed in more detail in Chapter 17.

## BREAST TOMOSYNTHESIS

Digital mammography has been demonstrated as providing equal or improved (for women with dense breasts) accuracy compared with screen-film mammography. Nevertheless, neither its sensitivity (probability of finding cancers when present) nor its specificity (probability of a negative result when cancer is not present) is 100%. One reason for this is the masking effect of tissue superposition that necessarily occurs in projection radiography. This masking effect can be reduced or avoided using 3-D X-ray imaging techniques. The principles of tomosynthesis are discussed in Chapter 10 and of computed tomography (CT) in Chapter 11. Dedicated breast imaging systems have been developed for both of these modalities. Both provide reconstructed planar images of sections of the breast.

Tomosynthesis images are acquired on a modified digital mammography system where the arm supporting the X-ray tube pivots about a point, while the compressed breast remains stationary. The detector may also pivot, depending upon the system design. Typically, a small number (9–25) of low dose projection images are obtained over a limited range of angles (±7° to ±30°) about the normal to the desired image plane.

The X-ray spectra used in tomosynthesis are generally higher in energy than those used in digital mammography (e.g. the W/Al spectrum shown inFig. 9.7). The X-ray tube may be moved in a continuous or discrete (‘step and shoot’) fashion; thus, short X-ray pulses with higher tube currents are used in tomosynthesis. As with mammography, the total acquisition time must be minimized to avoid image degradation due to patient motion.

Planar cross-sectional images are reconstructed from the projections using filtered back projection or an iterative reconstruction algorithm (see Chapter 11). It should be noted that the spatial resolution of tomosynthesis is anisotropic; tomosynthesis provides the highest resolution in-plane and relatively poor resolution between planes. As a result, the reconstructed voxels are generally non-isotropic, with a pixel size approximately equal to the size of the del and a reconstructed slice spacing that is typically 1 mm. Owing to the limited range of acquisition angles, the projection data for tomosynthesis do not form a complete set, so the reconstructed image is not a true 3-D representation of the breast anatomy. This results in artifacts, which are observed in images.

## BREAST CT

CT provides true tomographic images in which the voxels fairly accurately represent X-ray attenuation coefficients of the breast tissue. The breast CT systems that have been developed employ a cone-beam geometry and a flat panel area X-ray detector. Therefore, the data for all of the CT slices are acquired simultaneously. This allows rapid image acquisition, but causes the SPR to be much higher than would be the case in single-slice CT, especially as no grid is used. Voxels tend to be isotropic, but the pixel dimensions in the plane of the tomographic section are substantially larger than those provided with digital mammography or tomosynthesis. The current system designs provide a dedicated prone imaging table, which introduces challenges with respect to imaging tissue adjacent to the chest wall. Owing to the large number of projections, images are generally acquired at a much higher tube voltage (50–80 kV) than for mammography (~30 kV), in order to keep doses at an acceptable level. Nevertheless, the very low dose per projection can result in noisy images. A desirable feature of breast CT is that it can be performed without the need to compress the breast.

## COMPUTER-AIDEDAIDED DIAGNOSIS

The goal of computer-aided diagnosis (CAD) is to assist radiologists in detecting breast cancer, principally in screening mammography. CAD has the potential to be a cost-effective alternative to independent double reading bytwo radiologists, in that the CAD algorithm can be used to simulate the second radiologist. Double reading has been shown to increase the cancer detection rate, but it is not widely practiced because of costs and logistics. Thus, CAD has the potential to reduce the cancer miss rate, reduce the variability among radiologists, improve the consistency of a single radiologist and make radiologists more productive.

Most CAD schemes are designed using the paradigm shown in Fig. 9.13. A digital mammogram is used as the input for CAD. This can come from a full-field digital mammography system or from data acquired by digitizing a screen-film mammogram. The first step is to preprocess the image to segment the breast area from the non-breast area and to use image processing to emphasize lesions or certain features of lesions. For example, spatial filters can be used to make microcalcifications more prominent, or specialized non-linear filters can be used to highlight the spiculation associated with malignant masses.

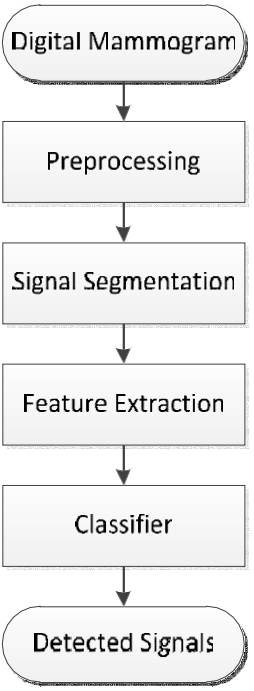


*FIG. 9.13. Conceptual outline of a CAD system.*

After the image has been preprocessed, potential lesions are identified. The simplest means is to 'threshold' the region containing the lesion (see Chapter 17), as both microcalcifications and masses appear brighter than their surrounding background. Once potential lesions have been identified, they are segmented from the image using various techniques (see Chapter 17). Because the borders of masses are often ill-defined or partially obscured by normal tissues of the breast, gradient-based methods are frequently more effective in these situations than grey-level-based methods.

To reduce the number of false detections, many features of the segmented detections are extracted from the image. Most features fall into one of three categories: intensity-based, morphology-based and texture-based. These features can be extracted from the greyscale image or from the image after it has undergone a mathematical transformation. A subset of these features is chosen for further analysis.

Once the final feature set has been chosen, the features are merged by a statistical classifier to differentiate actual lesions from false detections. Many different types of classifiers can be used, such as support vector machines,

artificial neural networks, k nearest neighbor and decision trees. Most classifiers have comparable performances and new classifiers are being actively researched.

CAD algorithms must be 'trained' using a set of mammograms for which 'truth' (i.e. the presence or absence of cancer) is known through biopsy resultsor subsequent patient follow-up. Truth data are also required to evaluate the performance of a CAD algorithm. Care must be taken in training and evaluating the classifier to avoid bias and to reduce the variance in the measured performance. To avoid a positive bias, cases used to train the classifier are not used to test the classifier. Increasing the number of training cases can improve the performance of the classifier and reduce the variance in the measured performance.

The results of the CAD algorithm are conveyed to the radiologist by means of an image annotated to show the computer detections. For screen film-based CAD systems, a low-resolution version of the image is either printed on paper or shown on a monitor. The locations of computer detected masses and clustered calcifications are shown using different symbols for the different lesions. For digital mammography systems, the CAD output can be annotated directly onto the radiologist's display workstation.

More recent research in CAD considers the combination of information from multiple images, either from a different view from the same examination, orfrom the same view from a previous exam. This approach more closely mimics how a radiologist reads a case, and it can improve the performance of a CAD scheme. Other CAD methods combine information from one or more images withclinical findings, as these data are also available and may influence the decisionsmade by clinicians. CAD may become particularly valuable in 3-D breast imaging, where the amount of image data to be considered is greatly increased. Here, CAD may be useful for automatic detection of microcalcifications in the large image set, allowing the radiologist to focus attention on more sophisticatedinterpretation tasks.

## STEREOTACTIC BIOPSY SYSTEMS

In mammography facilities that perform diagnostic procedures, digital systems are often used for guidance of stereotactic needle breast biopsy procedures. Stereotactic procedures are used to investigate suspicious mammographic or clinical findings without the need for surgical (excisional) biopsies, resulting in reduced patient risk, discomfort, and cost. In stereotactic biopsies, the gantry of a mammography machine is modified to allow angulated views of the breast (typically at ±15° from normal incidence) to be obtained (Fig. 9.14). From measurements obtained from these images, the 3-D location of a suspicious lesion is determined and a needle equipped with a spring-loaded cutting device can be

accurately placed in the breast to obtain tissue samples. These systems may use small format (5 cm × 5 cm) digital detectors or full field detectors. The image receptor can be located either in a conventional mammography unit or beneath a dedicated table, where the patient lies prone on the table with the breast pendant through an aperture in the table into the imaging region.

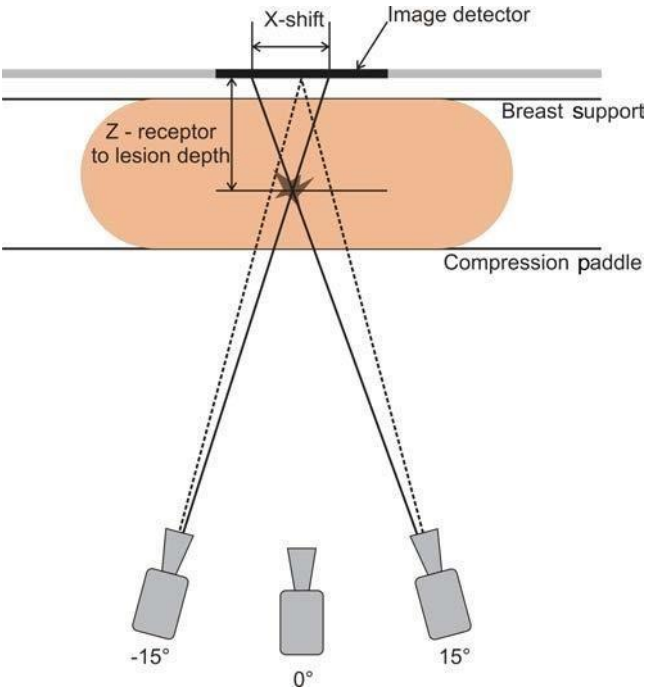


*FIG. 9.14. The geometry for stereotactic breast biopsy is shown. The X-ray tube is rotated about the breast to produce two views. The Z depth of an object can be determined by the lateral (X) displacement observed between the two views.*

## RADIATION DOSE

There are three dosimetric quantities used for mammography: (i) incident air kerma (IAK) ($K_i$), (ii) entrance surface air kerma ($K_e$) and (iii) mean dose to the glandular tissue of the breast, known as the mean glandular dose (MGD) ($D_G$). The MGD is the primary quantity of interest related to the risk of radiation-induced cancer in breast imaging.

Details of the calculation of MGD are given in Section 22.5.3.2. The MGD is calculated using factors obtained experimentally or by Monte Carlo radiation transport calculations, which convert from IAK to dose in a breast of specific composition and size. These conversion coefficients are tabulated in various publications, including IAEA Technical Reports Series No. 457 [9.1].

The MGD conversion coefficient increases with the mean energy of the X-ray spectrum. To produce an image of appropriate quality, every image receptor requires a specific quantity of X-ray energy to be transmitted by the breast and be absorbed by the receptor. As shown in Fig. 9.2, the attenuation of the breast decreases with increasing energy, so that the IAK required to obtain a specified absorbed energy in the receptor decreases accordingly. As energy increases, the required IAK falls more quickly than the conversion coefficient increases, so the net result is that the MGD falls with increasing energy.

Doses are not necessarily the same for all types of film or digital mammography system, especially for the latter where technologies differ quite markedly. In the Ontario (Canada) Breast Screening Program, in measurements made in 2009 and 2010, it was found on standard measurements with phantoms that the dose to each breast from the standard two-view examination was 3.2, 2.72 and 2.36 mGy for screen-film mammography, digital mammography with photostimulable phosphor cassettes (CR) and digital mammography with captive detectors (DR), respectively. In this survey, a system that used a photon-counting detector resulted in a dose that was only 41% of that required on average for imaging with screen-film technology.

The dose in mammography is also dependent on the size and composition of the breast, as well as the exposure settings selected. In screen-film mammography, where the goal is to maintain a target value for OD on the film, the IAK will increase as the thickness of the breast and the fraction of fibroglandular tissue (often referred to as density) increase. This will cause a corresponding increase in MGD. An increase in beam energy (tube voltage, choice of target material, beam filter) will mitigate against some of the dose increase. However, image contrast will be reduced and at some point this will become unacceptable.

In digital mammography, the goal is to achieve a target SDNR at the detector. Again, the dose will increase with breast thickness and breast density. However, with a digital system where contrast can be adjusted during image display, an acceptable compromise can be achieved at a higher energy than with screen film. This allows the advantage of a greater relative decrease of dose compared with film for large and/or dense breasts. Figure 9.15 shows the effect of breast thickness on the required dose, calculated using a theoretical model. Here, a constant SDNR is achieved at the detector. This requires increased dose for thick breasts and low energies where the breast is more opaque to X rays,but also for higher energies where the signal difference or contrast becomes less.

However, as previously mentioned, the increase with increasing energy is less than would be required for film mammography.

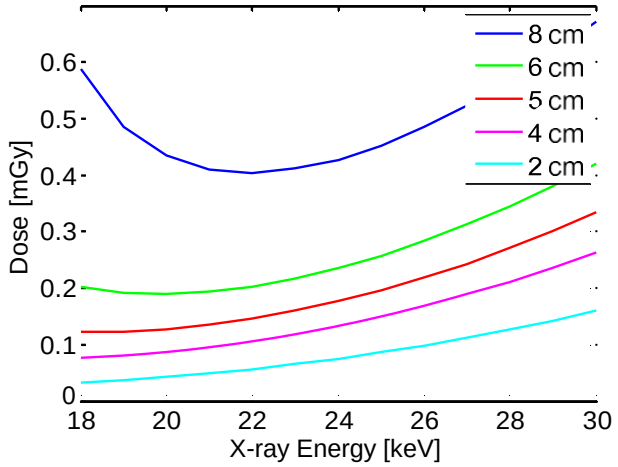


*FIG. 9.15. Calculated dose required to achieve a fixed SNR at the detector for breast thicknesses of 2–8 cm. These doses are somewhat low because the effect of scattered radiation on contrast and noise has not been considered and because the beam is assumed to be monoenergetic.*

There is a risk of cancer induction associated with the radiation doses received in mammography. Therefore, it is important to understand the magnitude of risk associated with the radiation dose delivered in mammography. The Biological Effects of Ionizing Radiation (BEIR) VII report critically examined data on doses and increased cancer incidence from several studied groups and allowed development of a radiation risk model for breast cancer [9.2]. The report provides a single model for all solid tumors, based on the work of Preston et al. [9.3]. For mammography, this model predicts the excess absolute risk (EAR) of cancer induction as:

$$\mathrm{EAR} = 10^{-3} e^{-0.05(A_x - 25)} \times \left(\frac{A}{50}\right)^{3.5} \quad (9.1)$$

for women of age, $A$, less than or equal to 50 and:

$$\mathrm{EAR} = 10^{-3} e^{-0.05(A_x - 25)} \times \frac{A}{} \quad (9.2)$$

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for women aged over 50 years. Here, EAR is the risk of the radiation induction of a cancer that will 'surface' in a 1-year period in a woman of age, *A*, per Gyof dose to the breasts at some earlier age, $A_x$, when the exposure occurred. Asan example, the risk for a woman aged 60 years from a dose to the breasts of 0.0024 Gy, previously received from mammograms at the age of 45 years is predicted to be $1.06 \times 10^{-6}$. Note that in this model, risk is linearly related tothe dose received and decreases with age at exposure. No explicit provision for latency is built into this model. However, the predicted risk does increase with attained age following the exposure, the rate of rise being slower after the age of50 years.

Integrated appropriately, this model can be useful in predicting the lifetimerisk following a single mammographic examination or from multiple exposures at different ages, as would occur in periodic screening. For example, in a screeningregimen that consists of annual mammography examinations from the ages of 40to 55 years, and biennial examination thereafter until the age of 74 years (i.e. 25 screenings) with a dose of 2.4 mGy to both breasts, it is estimated that in 100 000women, 56 radiation-induced cancers will be caused, resulting in 6.9 deaths and aloss of 88.5 women-years of life. For this calculation, a latency of ten years was
applied, i.e. EAR = 0 for $A < A_x + 10$. For these woman, earlier detection throughscreening would save 500 lives or 10 670 women-years, resulting in a benefit
to risk ratio of 72.5 (in lives) or 120.6 (in women-years). If the same diagnostic accuracy could be achieved at reduced radiation dose, the benefit to risk ratio would be even higher.