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Balancing Image Quality and Radiation Safety in Mammography: A Pilot Study of Detector Performance and Optimization Strategies

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ABSTRACT

Mammography represents a specialized diagnostic imaging technique that employs low-energy X-rays to generate high-resolution images of breast tissue for both diagnostic and screening applications. The methodology necessitates meticulous calibration of kilovoltage and tube current-time parameters, specifically customized to the unique thickness of the breast, to guarantee superior image quality while concurrently minimizing radiation exposure. The principal aim of mammography is the identification of low-contrast visible masses and calcifications, which may be as small as 100 microns in diameter, thereby demanding high spatial resolution in comparison to traditional X-ray imaging techniques. With the radiosensitive qualities of breast tissue in mind, it is necessary to keep radiation doses as low as is reasonably achievable (ALARA) to reduce the associated risk of cancer from X-ray exposure. This investigation aspires to offer a comprehensive analysis of the operational principles governing mammography apparatus and the diverse detectors utilised in mammographic imaging. Through the exploration of contemporary detector technologies and operational methodologies, this research endeavors to pinpoint opportunities for enhancing radiation protection within mammography systems. The objective is to elevate image quality while diminishing radiation exposure to breast tissue, thereby augmenting the overall safety and effectiveness of mammography as a diagnostic instrument.

Keywords: Mammography, Radiation protection, Detector technology, Operational principles, Breast imaging, Diagnostic radiology.

INTRODUCTION

Mammography is a radiological examination of the breast. It originated in 1913 when a German Surgeon, Albert Salomon, performed X-ray studies on excised breast tissue, correlating radiographic images with gross and microscopic anatomy. This work laid the foundation for mammography as a diagnostic tool. Breast cancer is a growing problem, and the occurrence of breast cancer increases with age. Mammography is the first choice for the diagnosis of breast cancer. Presently, it is still the most important breast imaging technique. X-ray mammography has also proved to be effective in reducing breast cancer mortality in a number of screening programmes (Zhou and Gordon, 1989; Hurley and Kaldor, 1992). The X-rays used in mammography are produced from molybdenum, rhodium, or tungsten, or a combination of any two of them as anode or target material. Most mammography tubes use beryllium exit windows between the evacuated tube and the atmosphere. The X-ray beam is produced when relatively large amounts of electrical energy are transferred from the cathode and collide with the target or anode material (Dance *et al.*, 2014). However, during the procedures and due to the use of low energy in imaging, breast sensitive tissue will be exposed to a significant radiation dose. One of the risks of mammography is the carcinogenic potential of the ionizing radiation (UNSCEAR, 2000). The X-ray machines used for mammograms today expose the breast to much less radiation than those used in the past. The X-rays do not go through tissue as easily as those used for routine chest X-rays or X-rays of the arms or legs, which also improves the image quality (American Cancer Society, 2013). The work is aimed at evaluating the performance of different mammography detector systems and exploring optimization strategies that maintain or improve image quality while minimizing patient radiation exposure.

Screening mammograms

These are X-ray examinations of the breasts that are used for women who have no breast symptoms or signs of breast cancer (such as a previous abnormal mammogram). The goal of a screening mammogram is to find breast cancer when it's too small to be felt by

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a woman or her doctor. Finding breast cancer early (before they have grown and spread) greatly improves a woman's chances for successful treatment. A screening mammogram usually takes 2 X-ray pictures (views) of each breast (Cranio-caudal view and Mediolateral oblique view), except in the case of women with large breasts, where more views are taken to see as much breast tissue as possible (ACR, 2003). Some women, such as those with large breasts, may need to have more pictures to see as much breast tissue as possible (American Cancer Society, 2013). It is widely known that breast cancer can be induced by high doses of ionizing radiation, such as X-rays, and the probability of induction is believed to be dependent on dose. Breast cancer induction decreases with increasing age at exposure; therefore, it is advisable to set the age range for screening programs at an older age. The American College of Radiology (ACR) recommends a baseline mammogram by age 40, twice yearly examinations between ages 40 and 50, and once yearly examinations after 50 years of age. For high-risk factor women, screening mammography can be implemented at an earlier age. Figures 1 and 2 show images of a mammogram at different views

Figure 1: Mediolateral oblique view (Akwo *et al.*, 2024)

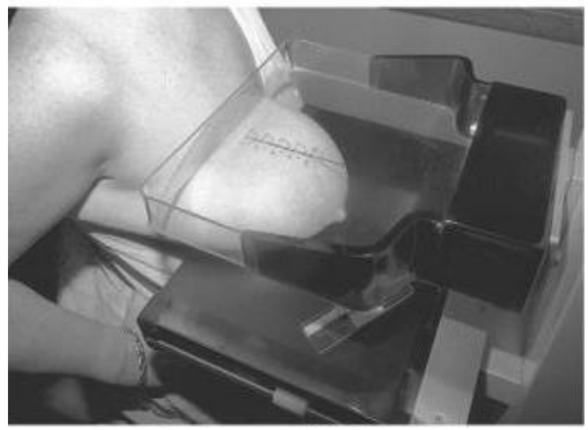

Figure 2: Cranio-caudal view (Akwo *et al.*, 2024)

Diagnostic mammography

Diagnostic mammography is used after suspicious (e.g. a lump or nipple discharge) results on a screening mammogram or after some signs of breast cancer alert the physician to check the tissue. During a diagnostic mammography, more X-ray images are taken, providing views of the breast from different angles. Magnification mammography may be undertaken to zoom into a specific area of the breast where there is a suspicion of an abnormality. This gives a better image of the tissue for more accurate diagnosis (Theerakul, 2014). Screen-film/conventional mammography and digital mammography are the two main mammography systems most commonly used for the detection and diagnosis of breast cancer.

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. These screens are used in conjunction with single-emulsion radiographic film, enclosed within a lightproof cassette. The formation of the X-ray image on the film process includes the X-ray beam emerging from the patient being converted into a pattern of visible light by the cassette's intensifying screen, and the visible image is then captured permanently on the film. Upon chemical development of the film with a processor, the emulsion that contains the latent X-ray image is converted into specks of metallic silver. The screen and film are arranged such that the X-rays 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 screens are often constructed with rare-earth phosphors such as gadolinium oxysulfide (Gd_2O_2S). X-rays that pass through the tissue are collected by the phosphor screens and converted to light. The film is positioned very close to the screen to capture the light photons; the image is then obtained by exposing the film. Films have significant disadvantages and limitations which include the lack of ability to detect small differences in contrast. The poorly exposed nature of images due to the film's stringent requirements for proper exposure, which leads to repeated exposures and reduced visibility of microcalcifications. There is also an increase in the examination time due to the chemical processing of film; they require large storage space in patient health records, and must be transported physically to the physician for viewing (Haus and Yaffe, 2000).

Digital Mammography

Digital mammography, which was introduced in early 2000, also uses X-rays to produce images of the breast, but unlike the screen-film system, there is a detector that converts the X-rays to digital images and they are stored directly in a computer. 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**.** There are currently two types of digital mammography systems: computed radiography (CR) which uses indirect digital detectors, and digital radiography (DR) which uses direct digital detectors. With a CR system, the X-rays transmitted through the breast are absorbed by the CR imaging plate, a Photostimulable storage phosphor (PSP). Absorbed X-ray energy corresponding to anatomical variations in the breast produces an electronic latent image on the PSP. The cassette is then removed from the mammography system and placed in a CR reader with a scanning laser beam that stimulates the release of light corresponding to the incident X-ray intensity. The light information is captured, converted to a digital signal, and displayed at the workstation. However, with the DR system, the X-ray signal is directly converted to a digital signal at the acquisition stand. No cassette is used. The image is displayed at the workstation shortly after it is acquired (Boissonnat *et al.*, 2023). Digital mammography systems offer a number of practical advantages and patient conveniences over conventional systems. Digital images are immediately available, therefore reduces the waiting time. The quality of the images can be evaluated as they are being taken; digital systems are fast, so patients spend less time in uncomfortable positions. Contrast of the images can be adjusted and sections of an image can be magnified after the mammogram is complete making it easier to see subtle differences between tissues. Digital images can easily be stored and retrieved and their transmission from one physician to another is quick and easy, multiple copies can easily be printed with the digital system (Chevalier *et al.*, 2012).

Computed Radiography mammography

The computed radiography (CR) operates like the screen-film mammography system. The CR detector which is in the form of a cassette with the image phosphor plate (typically made from barium fluorobromide BaFBr: Eu^{2+}) converts incident x-ray photons into a proportional number of electrons which could be trapped in the phosphor material to form a latent image. The latent image is extracted by translating the plate through a scanning red laser beam of 50- micron diameter. The laser energy stimulates the trapped electrons, which fall to the ground state by releasing photo stimulated luminescence energy in the form of blue light. This light is detected by photomultiplier tube which converts light energy into an electrical signal. The resultant signal is logarithmically amplified, digitized and processed for soft copy display using high resolution monitors. The imaging plate is erased by exposure to white light present in the CR image reader and thus the image plate can be reused. Mammographic images obtained using CR can yield image resolution of about 100 microns as seen in Figure 3, however, dual side reader yields a resolution of 50-microns pixel size. The advantages of using CR are that, standard mammographic cassette sizes in existing mammography equipment can be used to replace film screen radiography cassettes. The cost of CR system is much lower compared to other full field digital mammography systems (Cong *et al.*, 2019).

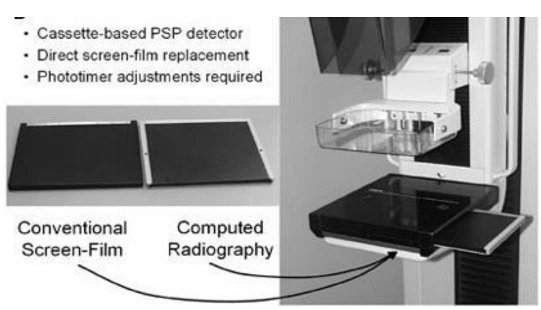

Figure 3: Computed Radiography mammography (Singh *et al*., 2011)

Slot scan digital mammography

A slot-scan mammography system, seen in Figure 4, consists of a scanning gantry and an enclosure for the X-ray power supply and other electrical components. In this system, a collimated narrow fan beam of X-rays scans synchronously across the full field of view, covering the entire breast tissue. The X-ray photons, after being transmitted through the breast tissue, fall onto thallium-activated caesium iodide scintillation material, which produces light. This light is channeled through an optic coupling to an array of charge-coupled device (CCD) detectors to form a digital image. The x-ray tube-detector assembly scans from right to left, continuously acquiring a signal. The detector is about 1cm wide in the scanning direction and consists of about 4 CCDs fused and is sufficiently long (22 cm) to include the entire area of the breast in the | anteroposterior direction. The CCD matrix consists of 400 x 2048 pixels with a field of view of 21 x 29 cm, producing images that are 4096 x 5625 pixels. The intrinsic pixel dimension is 25 microns, but a 2x2 bin is averaged to achieve 50 micron resolution at the detector plane for most image acquisitions (Cong *et al.*, 2019).

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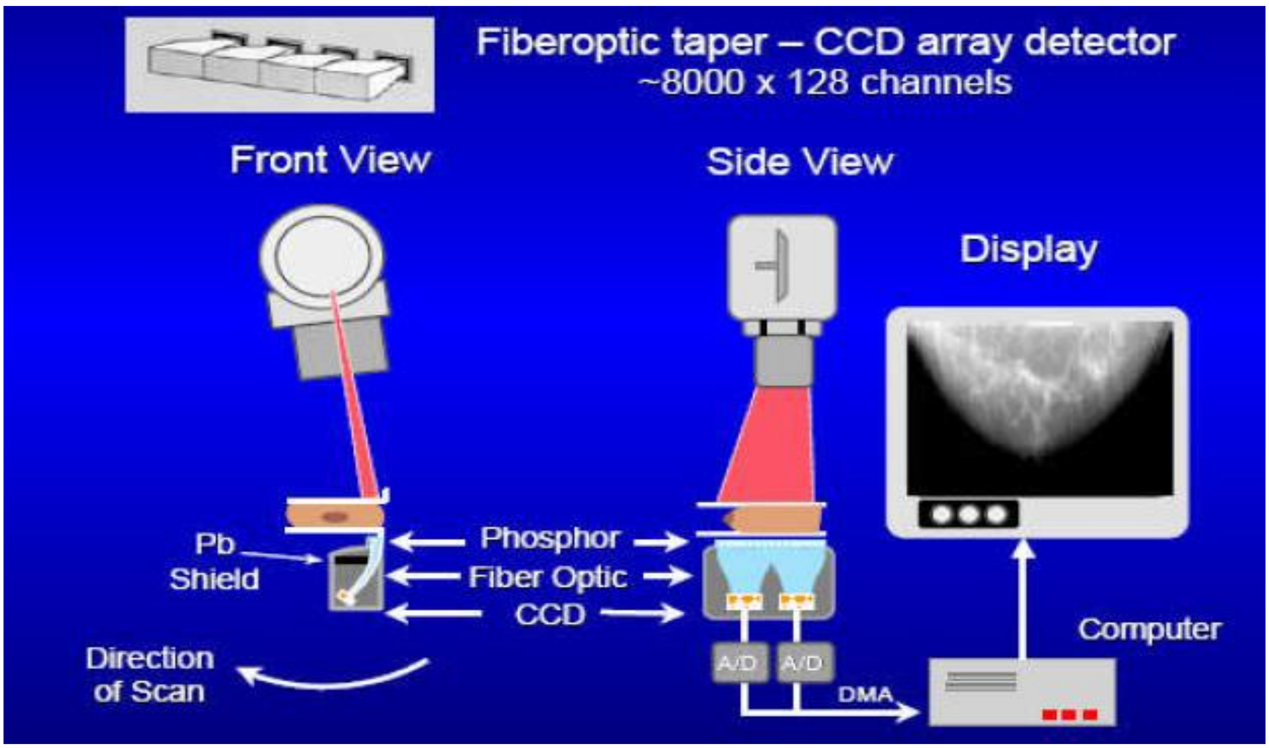

Figure 4: A slot-scan mammography system with accessories (Cong *et al.*, 2019).

The slot-scan system has a curved detector plate in order to allow the X-ray tube-detector array to sweep across the breast at the same distance with one uniform motion, with a typical scan time of 5 seconds. During the duration of the scan, the breast is immobilized in the compression paddle assembly. The advantage of this system is the elimination of the anti-scatter grid required for all other digital mammography devices, as the amount of scatter is significantly reduced by the small volume of tissue irradiated at any specific time. This leads to a significantly reduced breast dose. Slot-scan geometry requires high tube loading and necessitates the use of a tungsten-rhenium anode to withstand the higher heating that occurs with extended tube operation. There is, however, a choice of 3 filters, such as molybdenum, rhodium, or aluminum. Typically, a higher tube potential (31 kVp) is used compared to molybdenum or rhodium target X-ray tubes. Figure 5 shows a design of the slot scan mammography machine with a curved detector assembly (Elbakri *et al.,* 2005; Gennaro *et al.,* 2024)

Figure 5: A design of the slot scan mammography machine with a curved detector assembly (Elbakri *et al.,* 2005)

Indirect conversion with scintillation and thin film transistor

Basic principle: An indirect detector uses a scintillator to absorb and convert X-rays to light, and a photodiode array of amorphous silicon placed in each detector element to convert light to electric charge. In this detector, amorphous silicon (aSi) diode arrays are constructed from a matrix of aSi thin film transistor (TFT) arrays deposited on a glass substrate. Caesium iodide (CsI) scintillation crystals are vapor deposited on the aSi detector array. The CsI crystals grow in a needle-shaped light-pipe arrangement where the crystal structure internally reflects light to minimize light spread, thus allowing the phosphor to be relatively thick for good X-ray absorption efficiency and at the same time preserving spatial resolution. When the transmitted X-rays interact with the CsI crystal, they produce light. This light is converted into charge by- a sensitive aSi diode element which is integrated with TFT (acts as an on-off switch), a charge collector electrode to capture signals, and a capacitor to store signals produced from the X-ray converter. All elements are connected in a row and column matrix and using suitable algorithms, the image is processed to view on high resolution monitors. Currently available indirect conversion detectors include a single field of view of 19 x 23 cm, with a 100- micron detector element size. Although the 100-micron size seems to be large relative to conventional screen-film mammography (25 microns), the system produces good contrast resolution and reasonable spatial resolution (Elbakri *et al.,* 2005; Gennaro *et al.,* 2024).

Direct conversion and thin film transistor:

Basic principle: The X-ray photons are directly converted into electrical charges within the detector. The amorphous selenium used in a direct conversion detector is deposited directly onto the TFT substrate; there is no light conversion in this method. The direct detector uses amorphous selenium (aSe) semiconductor material placed between two electrodes. During operation, a high voltage is applied across the electrodes. As transmitted X-rays are absorbed in the selenium, electron-hole pairs (a hole is the positive charge carrier) are produced. The number of charge pairs is proportional to the incident x-ray fluence. The high voltage serves to separate electrons and the holes in order to avoid recombination, as well as to minimize the lateral spread of the like charges during transit to the opposite polarity electrode. The aSe direct conversion detector has excellent spatial resolution compared to other imaging detectors. These systems have a field of view of approximately 25 x 29 cm and have a pixel size of 70- microns. The limitations of this detector is its large storage space per examination since overall spatial resolution depends on the large number of small pixels, pixel size of the readout array and the type of digital sampling. The large detector is useful for imaging all breast sizes and presents no difficulty in positioning small breasts. The automatic exposure control senses the entire area of the breast and uses the area of densest portion of the breast to determine the appropriate exposure. This method of using the AEC is similar to the aSi diode array indirect conversion system (Hurley and Kaldor, 1992).

Mammography workstation: The monitor required for the mammography workstation should have a minimum pixel dimension of 2000 x 2500 (5 MP) with a luminance of above 600 cd/m2. In the absence of soft-copy reading using monitors, digital images can be printed using laser printers and read on high-luminance view boxes similar to viewing film images. This approach does not allow the user to benefit from the full range of post-processing options available on digital displays. The image processing options available on the work workstation include edge enhancement, adjustment of contrast and brightness, magnification, etc. Newer technologies include computer-aided diagnosis, which facilitates improved image reporting by radiologists. The other advances include breast tomosynthesis, dual energy imaging, digital subtraction mammography, and contrast-enhanced mammography (Hurley and Kaldor, 1992).

Components of Mammography Equipment

The mammography equipment consists of an X-ray tube and an image receptor mounted on opposite sides of a mechanical assembly, a filter, a breast compression plate, Automatic Exposure Control (AEC), source to image distance, grids, and magnification as shown in Figure 6. Since the breast will be imaged from different angles, the assembly can be rotated about a horizontal axis. Its elevation can also be adjusted to accommodate patients of various heights. The mammography system's geometry is arranged such that a vertical line from the focal spot of the X-ray source grazes the chest wall of the patient as depicted in Figure 6. This is to ensure that most, if not all, the tissue near the chest wall will be imaged.

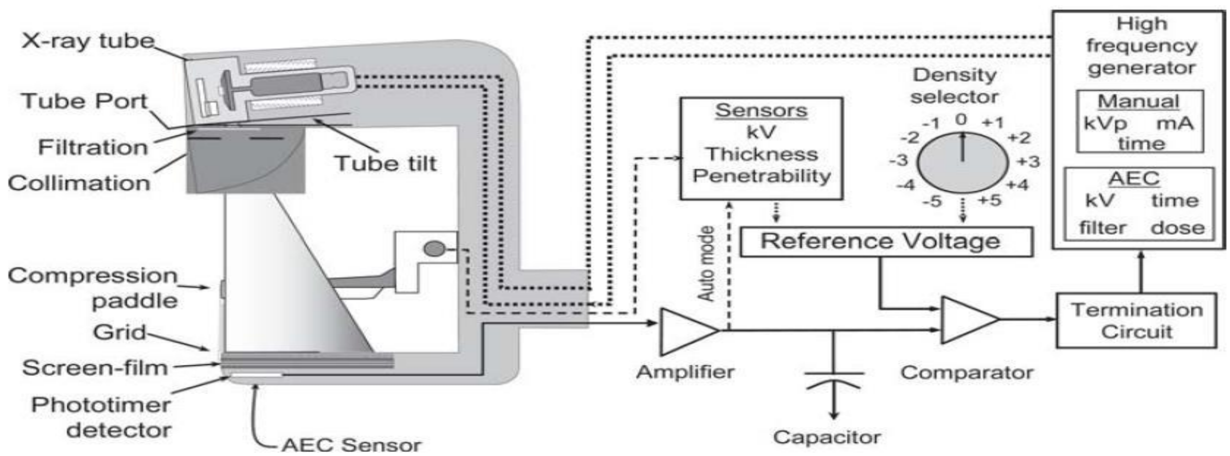

Figure 6: Parts of a mammography machine (Wase, 2024)

X-ray Tube

In mammography, the X-ray tube is usually operated at a kilovoltage of 25 to 32 kVp with molybdenum and rhodium as target materials in the anode. Both molybdenum and rhodium are used as target materials in the anode because they produce characteristic X-ray radiation at optimal energy levels. The mammography X-ray beam consists of bremsstrahlung and characteristic X-rays. Molybdenum has characteristic X-rays of 17.9 and 19.5 keV, while rhodium has characteristic X-rays of 20.2 and 22.7 keV (Bick and Diekmann, 2010). The slightly higher energy X-rays from rhodium provide better penetration of thicker or denser breasts. To reduce exposure times, typical X-ray tube currents are usually high, between 80 and 100 milliamperes (mA). Exposure times are usually about 1 second, but may be as long as 4 seconds for dense, thick breasts. In mammography, the X-ray tube is arranged such that the cathode side of the tube is adjacent to the patient's chest wall (anode heel effect), this is because the highest 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 (Dance *et al.*, 2014).

The Compression plate

A compression device is a plastic paddle used to flatten and immobilize the breast. Compression helps reduce motion blurring in the breast, separates structures within the breast, and decreases the thickness of breast tissue. This minimizes the amount of radiation required and the amount of scattered radiation reaching the image receptor. Ideally, the compression device is made of rigid, thin plastic and has a flat bottom surface that is parallel to the plane of the image receptor, with edges perpendicular to the plane of the image receptor to assist in moving breast tissue away from the chest wall and into the field of view. Adequate compression is essential for high-quality mammography. Compression reduces the thickness of tissue that must be penetrated by radiation, thereby reducing scattered radiation and increasing contrast, while reducing radiation exposure to the breast. Compression improves image sharpness by reducing the breast thickness, thereby minimizing focal spot blurring of structures in the image, and by minimizing patient motion. In addition, compression makes the thickness of the breast more uniform, resulting in more uniform image densities and in an image that may be easier to interpret. The displayed compressed breast thickness is often used to choose the technique factors, so this level of accuracy must be achieved (IAEA, 2014).

Filters

In mammography, molybdenum is the standard filter material. Filters as shown in Figure 7 are used to enhance contrast sensitivity. When a molybdenum filter is selected, it attenuates and blocks much of the bremsstrahlung spectrum above the energy of 20 keV. With the rhodium filter the k edge boundary is shifted to a higher energy (23.22 keV) so that the portion of the bremsstrahlung between 20keV and 23.22 keV is added to the x-ray beam. This makes the beam more penetrating than when using the molybdenum filter and provides some advantage when imaging larger or denser breast. In some systems, silver or tin is also used as filter material (Haus and Yaffe, 2000).

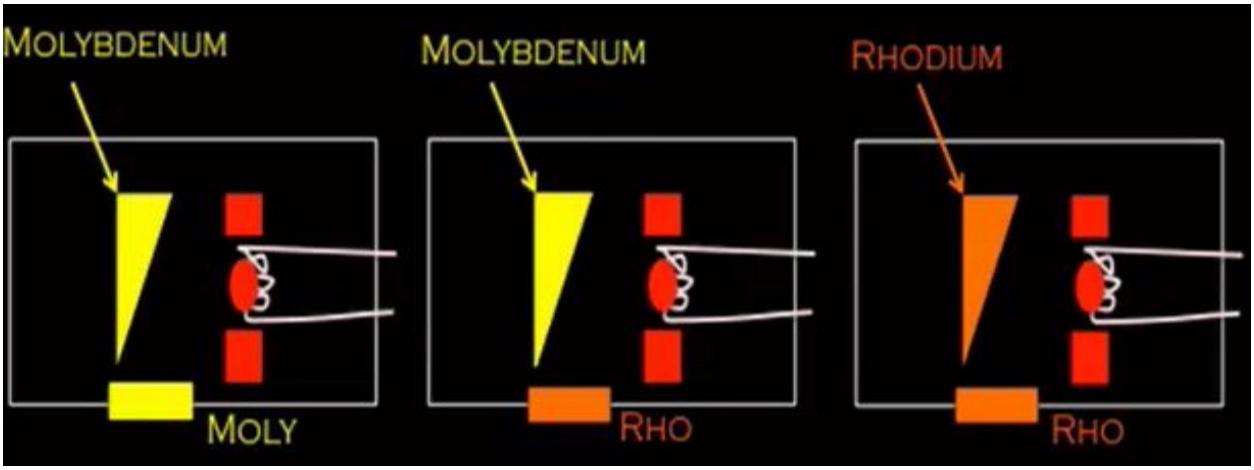

Figure 7: Target/ filter combinations in mammography (Haus and Yaffe, 2000).

Focal Spot:

The typical X-ray tube for mammography has two selectable focal spots, namely fine (0.1mm) and broad focus (0.3mm). The focal spots are generally smaller than those used for routine radiography because mammography requires images with minimal blurring and good visibility of detail to see the small calcifications. The fine focal spot is generally used for the magnification technique (Bick and Diekmann, 2010).

High-voltage generator

A near DC high voltage waveform with a minimal ripple not greater than that produced by a 6-pulse rectification system is required. The tube voltage should be accurate to ± 1 kV and reproducible to ± 0.5 kV. The tube current should be 100 mA or greater on large focus and 50 mA or greater on fine focus. The tube current exposure time product (mAs) range should be at least 5 to 800 mAs (Ciraj-Bjelac *et al.*, 2011).

Grids

Scattered radiation reduces contrast in all radiographic images. High contrast is especially important in mammography, where cancerous tissue and normal tissue appear quite similar since the densities of the tissues are almost the same. In the imaging of dense or large breasts, image contrast can be reduced by scattered radiation from the breast. The scatter-to-primary ratio for a breast of average size and density can be 0.6 or greater (Bick and Diekmann, 2010). Reducing the effects of scattered radiation has been achieved through the use of scatter-absorbing grids. To offset the reduction in X-rays caused by the grids shown in Figures 8 and 9, an increase in the mAs to about 2 to 2.5 times the value for non-grid techniques is required. The increase in mAs increases patient dose. The grid should have a high transmittance of primary radiation; this is to avoid an excessive increase in patient dose. For this reason, carbon-fibre covers and fibre-interspace materials are often used.

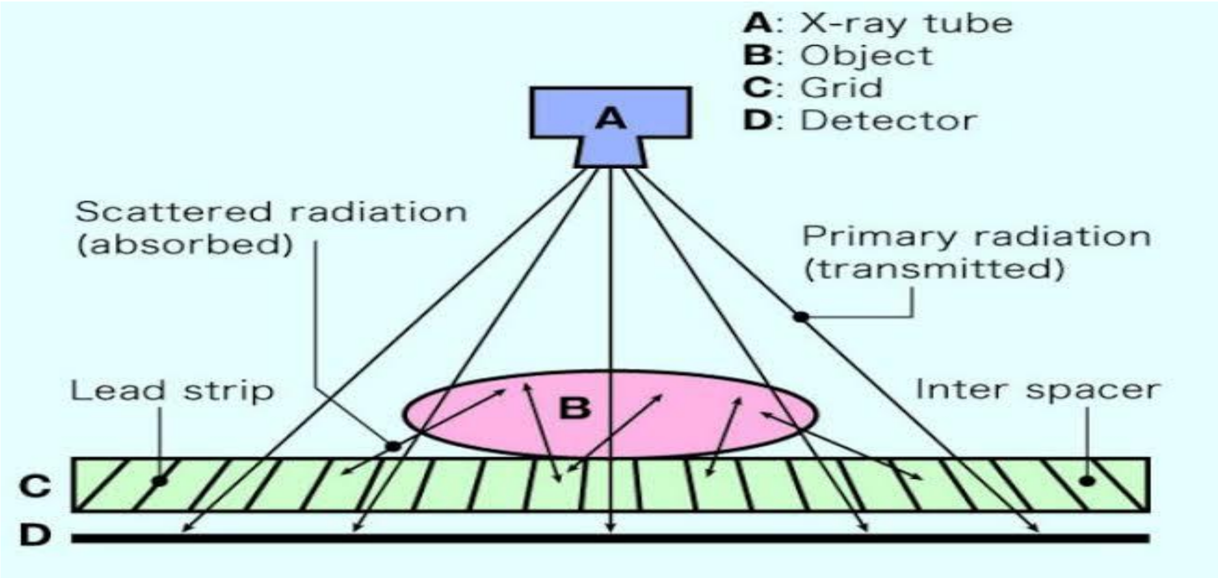

Figure 8: Description of grid (Bick and Diekmann, 2010)

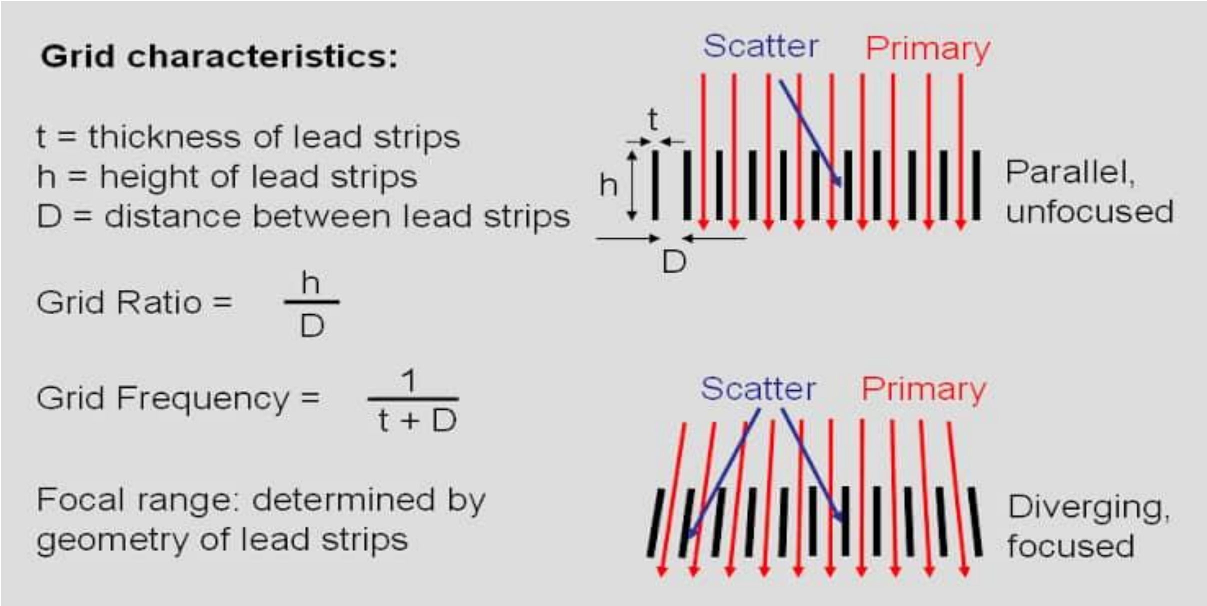

Figure 9: Grid characteristics (Bick and Diekmann, 2010)

Automatic exposure control (AEC)

In the early days, the exposure parameters such as kV, mAs, film screen combination, optical density, target material and filter material were manually set by the technologist after evaluating the breast thickness and density. Use of the AEC system automatically selects the exposure parameter depending upon the thickness and density of the breast. The AEC detector is located below the image receptor and is movable to cover different anatomical sites on the breast. The system is capable of being calibrated for different film-screen combinations provided by various vendors. The AEC should be reliable and consistent in operation, ideally with a variation of less than ± 0.2 in optical density over the appropriate range of tube voltage and breast thickness (Ciraj-Bjelac *et al.,* 2011).

Image Receptor:

Both film-screen and digital receptors are used in mammography imaging. Digital mammography consists in the replacement of the conventional screen-film analog detector with a digital detector (Rofena *et al.*, 2023).

Film-screen mammography (traditional system)

In a film screen mammography system depicted in figure 10, the entire imaging process is captured, displayed, and archived with a single medium, which is the radiographic film. Earlier days, non-screen cassettes with slow speed films were used for mammography in order to obtain finer details. However, over the years, single side emulsion coated films with single sided intensifying screen (preferably absorptive base) are being used in order to reduce the radiation dose and obtain diagnostic quality images. The single side emulsion films can be identified with a notch on the right-hand side of the film to identify while loading in the dark room (Mori *et al.*, 2023).

The inherent advantage of the film is its high spatial resolution (up to 20-line pairs per millimeter), which can demonstrate fine speculations and micro calcifications; high contrast details which allows visualization of subtle differences among soft-tissue densities. However, it requires high-luminance view boxes to view and display the films. The film-screen cassettes are also available in multiple sizes which enables to imaging breasts of different sizes. In addition, film acts as an efficient medium for long-term storage with low cost, and overall the advantages make the film as the standard of reference in mammography. Despite these advantages, there are a number of inherent limitations with film due to the limited dynamic range. The system should be optimized for the dense part of the breast where the automatic exposure control sensor is to be positioned in order to get a uniform density of film. An overexposure may lead to film darkening and is not fit to be used for diagnostic purposes. Hence, the mammography systems are tuned for films supplied by different vendors.

Other factors that play a role in the imaging of mammographic images are the temperature and the strength of the film processing solutions present in the dark room. Image artifacts also limit the use of film-screen mammography (Mackenzie *et al.*, 2025) shadow is significant especially at the low X-ray energies used in mammography. When it is placed behind, the amount of X-ray flux reaching the controlling detector is influenced strongly by absorption in the image receptor, which is highly dependent on the energy of the X-ray photons emerging from the patient. The AEC should be capable of maintaining optical density within ± 0.15 OD as the peak kilovoltage is varied from 25kVp to 35 kVp and as the breast thickness is varied from 2.5 to 8 cm for each technique. A range of density selections are available. Also, there are adjustments to provide proper compensation for different film-screen combinations. Exposure is terminated when the radiation dose received by the detector reaches a pre-determined level, which corresponds to the desired optical density. For many mammography units, the position of the radiation detector can be varied between two or more predetermined positions to facilitate the exposure of breasts of different size and density (Assiamah, 2004)

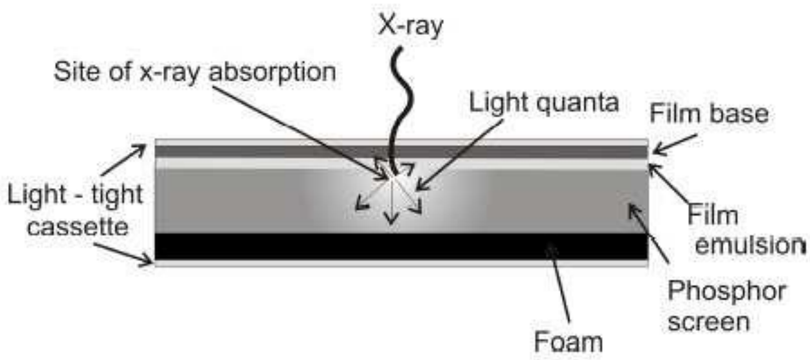

Figure 7: Configuration for a mammographic screen-film image receptor. A single-emulsion radiographic film is held in close contact with a fluorescent screen in a light-proof cassette (Dimakopoulou *et al.*, 2006).

Magnification

A magnification is the enlargement of a suspicious area of a mammogram. A magnification view is performed to evaluate and count microcalcifications and assess the borders and tissue structures of the suspicious area or mass by using a magnification device, which brings the breast away from the film plate and closer to the x-ray source. This allows the acquisition of magnified images of the region of interest (Gennaro *et al.,* 2024).

Source-to-image distance

This is the entire distance of the X-ray beam from the focal spot (source) to the image receptor, the greater the distance, the lesser the occurrence of geometric blurring. This affects the focal spot size, which in turn affects the size of the object being imaged. The larger the SID, the larger the field size. With a larger source to image distance (65cm), more beam penetrability is required, which in turn increases the heel effect (Gennaro *et al.,* 2024).

Mammography Dosimetry

There is a significant risk of radiation-induced carcinogenesis associated with mammography. The determination of mean glandular breast dose forms an important part of the quality control of mammography imaging systems since it gives an indication of the performance of the imaging system as well as estimating risk to the patient. As a result of the fact that breast cancer almost always arises from the glandular tissue of the breast, and also based on the assumption that it is the glandular tissue in the breast that is most sensitive to radiation effects. The mean or average radiation absorbed dose of the glandular tissue is used to measure the radiation risk associated with mammography (IAEA, 2014). The mean glandular dose is defined as the average dose to the glandular tissue. Because of the difficulty of estimating mean glandular dose (MGD) directly, the entrance air kerma, without backscatter, at the upper surface of the breast is determined and the MGD is estimated by multiplying by appropriate conversion factors (Dance *et al.,* 2000).

The mean glandular dose is calculated using equation

$$MGD = K.g.c.s \qquad (1)$$

where K is the incident air kerma (without back scatter) at the upper surface of the breast, g is the incident air kerma to mean glandular dose conversion factor (g-factor), c corrects for any difference in breast composition from 50% glandularity and the factor s corrects for any difference due to the use of a different X-ray spectrum. The conversion factors g, c and s are extrapolated from the work of Dance et al. (2000; 2009; 1990).

CONCLUSION

The preliminary analysis highlights the essential requirement of attaining a thoughtful equilibrium between imaging quality and radiation safety in mammography through systematic evaluation of detector effectiveness and investigation of specific optimization techniques. Our research findings clarify that it is indeed feasible to maintain elevated diagnostic image quality while adhering to established radiation dose parameters through the careful selection of detector technologies and the application of advanced imaging modalities. Optimization methodologies such as customized exposure configurations, noise-reduction algorithms, and detector calibration have been demonstrated to enhance image quality without exacerbating radiation exposure. These findings underscore the potential for the refinement of clinical protocols in a manner that bolsters both patient safety and diagnostic performance. Further comprehensive investigations are recommended to validate these methodologies and facilitate their integration into conventional mammographic practice.

RECOMMENDATIONS

1. Optimize exposure parameters based on breast thickness and density by adjustment of tube voltage(kVp) and tube current(mAs) using automatic exposure control (AEC) systems to minimize dose without compromising image quality. Establish breast-thickness- specific protocols that guarantee adequate contrast while keeping the average glandular dose (AGD) within acceptable limits.
2. Estimate and select high - performance detectors with high quantum efficiency to maintain - image quality at lower radiation doses.
3. Normalize quality control procedures and calibration of imaging systems, including- phantom testing, to ensure consistent image quality as well as routine training of staff to adhere to optimized protocols to ensure reduction in repeat imaging measures.
4. Use feedback from radiologists to fine-tune protocols before broader implementation

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