Optimisation of Qualitative and Quantitative Assessment of Images in Digital Mammography

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To my mum who was continually praying for me. To my wife who was always there for me.
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Abstract

The main aim of the breast screening programme (BSP) is to provide early detection of breast lesions permitting efficient treatment. At present the BSP is technically run using conventional screen film (S/F) X-ray mammography, which suffers from a number of inherent limitations. Most of these limitations could be overcome by the use of digital mammography technology. Such a technology has to, however, show image performance that is as good as or better than that for S/F systems. The performance of a digital mammography system can be quantified and compared in a number of different ways. These include contrast detail and contrast-to-noise ratio (CNR) measurements. In practice, various limitations and problems have, however, become apparent in applying these measurements. For the CNR measurements the main problems are the presence of the uncorrected heel effect and the effect of the size of region of interest (ROI). The contrast detail analysis, using human observers, is time consuming and suffers from the presence of significant inter-observer error. These can be solved by using an “automatic observer”. One limitation of using the automated approach is that the relationship between automated and human observer scoring was not fully explored across the wide variety of systems and circumstances encountered in practice. An alternative approach would be to predict contrast detail response from measurements of modulation transfer function (MTF) and detection quantum efficiency (DQE) using a model of the imaging process. In this thesis, the procedure of measuring CNR was revised and the use of an automatic approach for contrast detail measurements was further examined, using different modalities of digital mammography. The contrast detail performance was then analysed across a range of doses for a wide range of clinically used digital mammography systems, using the human results predicted from the automated measurements. MTF and DQE were also measured for the detectors used in these systems. A simple signal-matched noise-integration model was then adopted to theoretically predict the contrast detail response of these systems. The most remarkable findings of this thesis are as follows: the use of multiple small ROIs led to CNR results that were essentially the same as if a heel effect correction had been applied; the automated measurements can be used to predict the threshold contrast for a typical observer; an encouragingly good level of agreement was found between the experimental contrast detail data and theoretical predictions. Finally, image performance of promising hybrid pixel semiconductor detectors, not commercially available, was also evaluated with the aid of Monte Carlo simulation, for application to digital mammography.
Chapter 1

Introduction

Breast cancer is the most common and most feared malignancy in women and the second leading cause of cancer death in developed countries. Moreover, the incidence of breast cancer continues to rise [1]. Nearly, 45,000 positive cases of breast cancer are registered annually in the UK i.e. about 144/100,000 of female population [2]. 55.5% of these cases die each year [3]. With this high mortality rate the only way of prevention from death caused by breast cancer is to detect the cancer in its earliest stages of development where current methods of treatment are very efficient [4]. Early detection of breast cancer is performed by running a breast screening programme (BSP) which results in reduction of the mortality rate and in improving patient prognosis [5]. For instance, the NHS breast screening programme in England saves around 1,400 lives each year [2].

X-ray mammography which exists as a useful tool for initial detection of breast cancer has become universally approved. It offers several advantages over all other techniques used for this purpose. In other words, it has become the gold standard method. However, the use of conventional screen-film (S/F) combination systems is accompanied by a number of inherent limitations in detecting tumours. The narrow dynamic range of the film, film processing problems and inflexibility in image postprocessing are the most obvious limitations of the cassette film technique. Moreover, current S/F systems have a sensitivity of 60 – 80% and have a specificity which is very low [6,7] (40% maximum) [7]. In fact, S/F mammography has a quite acceptable sensitivity to detect breast cancer in a medium density breast. In a high density breast, however, this sensitivity is insufficient [8]. It has been estimated that about 50% of the screening population have high density breasts for which women are at increased relative risk of developing breast cancer [9]. This would affect the
probability of cancer detection in particular at the early stage resulting in an increase of the false negative diagnosis.

Digital mammography, in contrast, offers several potential advantages including wider dynamic range and high image contrast which result in improving the sensitivity and specificity of breast cancer detection [10]. In addition, the possibility of decoupling the processes of image acquisition, storage, processing and display, and capability of using computer aided detection (CAD) and picture archiving and communication system (PACS) provide more potential for the use of digital mammography [1].

Recently, there have been various digital mammographic modalities developed to address the limitations of S/F combination systems and hence improve image quality and reduce received dose. For more than a decade, scintillator-based charge coupled devices (CCD), photostimulable phosphor (PSP) based computed radiography (CR) and flat panel imagers (FPI) based on caesium-iodide (CsI) scintillators combined with an amorphous silicon (a-Si) photodiode array digital mammography systems have become available. These are referred to as indirect systems which are, however, less efficient compared to the systems which use direct conversion detector, due to the intermediate step of the conversion of X-rays to light photons before finally converted into an electronic signal that can be digitised [11]. Direct conversion systems, in which the incident X-ray photons are directly converted into electron – hole (e-h) pairs, have become commercially available within the last few years. Theses systems are based on FPI coated with a photoconductive layer of amorphous selenium (a-Se) and thin-film transistor (TFT) matrix. In addition, other direct conversion solid-state semiconductor detectors such as cadmium telluride (CdTe), CdZnTe (CZT), gallium arsenide (GaAs) and lead iodide (PbI2) have been proposed for application to digital mammography [11,12]. These materials are ideal for mammography detectors, because they have high X-ray absorption efficiency and extremely high intrinsic spatial resolution and low noise [13].

All the above systems are based on energy integrated technology in which the electronic signals from all X-ray photon interactions in a single pixel are usually integrated and then digitised to form the final pixel value. This technology, however, suffers from a degradation of signal-to-noise ratio (SNR) due to the fluctuation in the
amount of light (in the indirect detectors) and numbers of charges collected at the output electrodes. A possible solution to this limitation is the use of the photon counting technique which has been recently launched in a scanned slot (or multi-slit) geometry. The technique is based on a solid state silicon detector and operated with an electronic threshold method by which signals below a preset threshold level are not included in the image resulting in improved SNR [14].

In fact, digitisation of radiography was successfully performed a long time ago [15,16]. It was, however, argued that the technology is most difficult and challenging for application to mammography [17] and there are several on-going debates about mammography going digital. A compromise has to be reached between image contrast and spatial resolution before deciding whether to adopt a particular system. This means that image performance of digital systems has to be as good as or better than that for S/F systems.

The performance of a digital mammography system can be quantified and compared in a number of different ways. European and UK guidelines for quality control in digital mammography define procedures for measuring image quality in terms of contrast detail detectability, based on readings of images of the contrast detail mammography (CDMAM) test object by human observers [18,19]. The guidelines also define a procedure for measuring contrast-to-noise ratio (CNR) using a 0.2 mm thickness of aluminium with different thicknesses of Plexiglas (PMMA). Physical Measurements such as modulation transfer function (MTF), normalised noise power spectrum (NNPS), noise equivalent quantum (NEQ) and detection quantum efficiency (DQE) are well established for assessing detector performance of an imaging system [20]. A protocol for measuring these imaging parameters of the detectors used in digital mammography was recently published [21].

The aim of this thesis is to further investigate the applicability of the current UK and European assessment procedures for digital mammography. The difficulties, limitations and problems associated with these evaluation procedures are discussed in detail. Also appropriate solutions and alternative methods are proposed. The performance of a wide range of the clinically used digital mammography systems are then evaluated and compared. However, the ultimate goal is to develop a reliable method of assessing digital mammography systems against European Guidelines.
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The thesis is outlined as follows. Chapter 2 gives a general background of the X-ray imaging, the principles of mammography and the appropriate image quality parameters. In chapter 3 the conventional S/F mammography and the current approaches of digital mammography will be reviewed and compared in terms of image quality and dose efficiency. In the following chapter, chapter 4, the procedure defined by the European guidelines for measuring CNR will be evaluated. The impact of some problems, such as heel effect and size of region of interest (ROI), in determining CNR will be investigated for digital mammography. Possible solutions will be then examined and presented. In chapter 5, the methods used to determine the contrast detail detectability will be further studied using a large number of human readers and a large population of digital mammograms. The automated approach of reading the CDMAM images will be also used and compared to the human readings. The reproducibility of automatic reading and its ratio to the human observer will be then obtained using different numbers of images and for different types of systems. Chapter 6 aims to investigate the effect of several clinical conditions on the determination of MTF for digital mammography systems and an optimal method will be proposed.

In chapter 7, the thesis is now moving into the main part of this project by considering whether the performance evaluated by subjective measurements (contrast detail analysis) is correlated with that expected from the physical (objective) measurements. The possibility of using the objective results to predict the subjective measurements will be also investigated using a theoretical model of the imaging process. In addition, chapter 7 will include a full evaluation of the performance of a wide range of the currently available digital mammography units by comparing their physical detector characteristics. Moreover, the contrast detail performance of these systems will be analysed across a range of doses and present the results as the dose required to achieve the standards in the European guidelines at different detail sizes.

Chapter 8 contains evaluation of the image performance of a variety of hybrid pixel semiconductor detectors for digital mammography application. This technology is not yet commercially available and they are still under development. Therefore the image performance of these detectors will be evaluated with the aid of Monte Carlo (MC) technique.
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While the last chapter, chapter 9, includes the conclusion of the whole thesis and suggestions for future work, the appendices contain all the source codes of the software programmes written by the author and used in this project. These are the MC, Matlab, and Visual Basic programmes. In addition a C++ programme was developed by Jon Denne, Computing Section, Medical Physics Department, Royal Surrey County Hospital, Guildford. The programme was solely produced for the purpose of the work made in Chapter 5.

References

1. E. D. Pisano, M. J. Yaffe, and C. M. Kuzmiak, Digital Mammography, Lippincott Williams & Wilking, USA, 2004
CHAPTER 1. INTRODUCTION


Chapter 2

Background

To study any digital medical imaging system, there is a need to understand the many concepts supporting such an activity. The interaction of photons with the detection material plays a major role in a medical imaging. The digital imager along with its pixel size and pitch is the key element in image production and performance. This has to be then evaluated using some important parameters. In this chapter, the production of X-rays and their interactions with matter, the mammography system and the essential assessment parameters will be discussed.

2.1 X-ray Production

The main component of an X-ray unit is the vacuum X-ray tube. The high voltage circuit in the tube consists of a positive pole, the anode, and a negative pole, the cathode. Diagnostic X-rays are produced when accelerated electrons, emitted from a heated filament in the cathode by the process called thermionic emission, strike the anode material. As a result, the kinetic energy of the electrons (99%) is transferred to the target atoms constituting unwanted heat. The rest of their energy (1% or less) is converted into X-rays [1]. The efficiency of X-rays production depends entirely on tube current, i.e. the number of electrons which flow from the cathode to the anode of the X-ray tube, and on tube voltage, i.e. the maximum tube kVp [2]. Typically, 0.5 – 1000 mA and 30 – 150 kVp are the tube current and voltage, respectively. In order to deal with the excessive heat generated in this process, a rapid rotating anode technique, copper (for rapid heat transfer) and oil (for cooling) are generally employed [3,4].

An electron from the filament has two probabilities in respect of collision with target material. In the first, the accelerated electron penetrates deep into the target atom and
collides with an orbital electron (more likely in the K-shell) which is then ejected from the atom creating a vacancy. An electron from an outer shell (most probably L-shell) subsequently falls to fill the vacancy in the K-shell with the emission of an x-ray whose energy is characteristic of the target atom. This energy is the difference in the binding energies of the two shells, \( E_K - E_L \) and the single x-ray emitted is called characteristic x-ray \( (K_a) \). Less likely, an electron from the M-shell might fill the hole created in the K-shell emitting an x-ray \( (K_B) \) with a relative higher energy \( (E_K - E_M) \). Fig. 2.1 shows the \( K_a \) and \( K_B \) characteristic x-rays as sharp lines on the X-ray spectrum obtained from a molybdenum (Mo) target with energies of 17.5 and 19.5 keV and intensities of 0.84 and 0.16 [5].

![Bremsstrahlung and characteristic x-rays](image)

Fig. 2.1. An example of X-ray energy spectrum obtained using a Molybdenum target material and 30kVp tube voltage. Reproduced from [6]

The second probability occurs when the bombarding electron approaches close to the positive nucleus resulting in a deflection by which some or all of its kinetic energy is lost. According to the energy conservation law, the lost energy is converted into a single photon of X-rays, known as braking radiation or bremsstrahlung. These X-rays can be emitted in any direction with a continuous range of energies up to the maximum value \( E_{max} \) equal to the tube voltage. The bremsstrahlung radiations form a continuous spectrum and the characteristic x-rays are superimposed, as sharp lines, on that spectrum as shown in Fig. 2.1. At energies typically used in mammography systems (28 keV), about 71% - 81% of the X-rays produced are bremsstrahlung [3,5].


2.2 Heel Effect

The angled anode design of an X-ray tube causes a slight reduction in X-ray intensity on the anode direction of the irradiated area of the detector. This phenomenon is called the *heel effect* which is the main source of the background in-homogeneities in X-ray images. As illustrated in Fig. 2.2, X-rays travelling toward the anode direction of the image field (side A in figure 2.2) are attenuated more than those travelling toward the cathode direction (side B figure 2.2) producing a non-uniform image. Fig. 2.3 shows an example of a non-uniform image obtained, by a mammography system, using a 45 mm Plexiglas (PMMA). Fig. 2.4 shows the profile measured across the image in Fig. 2.3 demonstrating the presence of low frequency components (non-uniformity) of the image due to the heel effect.

Fig. 2.2 illustrates the heel effect phenomenon

![Diagram of heel effect](image)

Fig. 2.3 illustrates the non-uniformity due to the heel effect of a mammography image. The double lines indicate the transverse profile, measured across the image, used to obtain Fig. 2.4
Fig. 2.4 shows the profile measured across the image in Fig. 2.3, demonstrating the presence of the low frequency trends, i.e., the variation in pixel value between the anode side (pixel number zero) to the cathode side (pixel number 3311) of the image due to the heel effect. The high frequency component, which is the statistical variation in pixel value between the pixels, is also demonstrated.

2.3 Interactions of photons with Matter

In the range of diagnostic X-ray energy, the predominant interactions of photons with matter are photoelectric absorption, and coherent and incoherent scattering.

2.3.1 Photoelectric Absorption

Photoelectric absorption occurs when a photon interacts with a tightly bound electron, of an absorber atom, whose binding energy is equal to or less than the energy of the incident photon. As a result, the incident photon completely disappears in the atom and in its place, a photoelectron, whose energy is the difference between the energy of the incoming photon and the binding energy of the electron with which it interacts, is ejected by the atom. This creates a vacancy in the bound shell of the absorber atom. This vacancy is quickly filled by a free electron from other shells of the atom, generating one or more characteristic x-ray photons.

The photoelectric absorption is the predominant mechanism of interaction for photons of relatively low energy $E$, and for absorber materials of high atomic ($Z$) number.
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However, the probability of photoelectric absorption per atom of the absorber can be approximately expressed [7], as

\[ P = \frac{Z^n}{E^{3.5}} \]

where \( n \) varies between 4 and 5 over the photon energy region of interest [7].

2.3.2 Incoherent Scattering (Compton Effect)

Compton scattering takes place when an incident photon interacts with a free electron in an absorbing material. As a result, the incoming photon is scattered at any angle with respect to its original direction. The photon transfers a fraction of its energy to the electron with which it interacts. Depending on the angle of scattering, the energy transferred to the electron varies from zero to a large portion of the photon energy.

The probability of Compton interaction per atom depends on the number of free electrons available in the absorbing material and therefore increases linearly with \( Z \) number of the absorber. The interaction probability, however, falls off gradually with increasing photon energy. The theoretical cross section for this reaction was derived by Klein and Nishina and a full description of this is found elsewhere [7].

Compton scattering is the predominant process across a wide range of diagnostic X-ray energies in particular when soft tissue is exposed. Photoelectric absorption, on the other hand, is the predominant interaction for X-ray films, screens and other X-ray detectors (e.g. digital X-ray detectors), due to the high \( Z \) number of these materials. For example, when an amorphous selenium (a-Se) flat panel is used as a digital X-ray detector, the ratio between the photoelectric and Compton attenuation coefficients is 482 to one [8]. However, in the range of mammography energy (typically 20 – 30 keV) the predominant interaction is Compton scattering. This holds true if we assume that breast is comprised only of soft tissues. In the presence of calcifications, photoelectric absorption might, however, be more important than the Compton scattering. This is clearly demonstrated in Fig. 2.5 where photoelectric (\( \gamma/\rho \)) and Compton (\( \sigma/\rho \)) mass attenuation coefficients of bone and soft tissue are plotted as a function of energy. The photoelectric mass attenuation coefficient of CsI(Tl), which is the typical detection material used for the S/F cassettes of conventional...
mammography systems and for the amorphous silicon (a-Si) detectors in digital mammography systems, is also shown for comparison.

### 2.3.3 Coherent Scattering (Rayleigh Scattering)

Rayleigh scattering occurs when an incoming photon interacts coherently with all the electrons of an absorber atom, i.e. with the entire atom rather than with an individual electron as in the Compton effect. After interaction, the atom is neither excited nor ionised and the scattered photon retains its original energy. This means that Rayleigh scattering effectively contributes nothing to the transfer of photon energy to matter [9]. Consequently, this mode of interaction is often given limited attention in basic discussions of photon interactions [7], and also in diagnostic radiology [9,10]. The probability of coherent scattering is significant only for low photon energies and is most important in high-Z absorbers. In general, it is negligible for photon energies greater than about 100 keV [7,10] in soft tissue [10]. However, Rayleigh scattering accounts for approximately 12% of X-ray interactions at the range of energies used in mammography (i.e. 20 – 30 keV) [11].

![Diagram of mass attenuation coefficients](image)

**Fig. 2.5.** The photoelectric ($\sigma/p$) and Compton ($\sigma/p$) mass attenuation coefficients of bone, soft tissue and CsI(Tl) as a function of energy. The blue lines indicate the typical range of mammography energy.

Note, the effective $Z$ number for soft tissue, bone and CsI(Tl) are: 7.4, 13 and 52, respectively.

Adapted from [3] with minor modifications.
2.4 Principles of Mammography

2.4.1 Composition of the Female Breast

The female breast consists of glandular, adipose (fatty) and fibrous tissues positioned over the pectoral muscles of the chest wall, as shown in Fig.2.6. In infancy, the breast is composed mainly of adipose tissue. The fibroglandular tissue begins to develop at puberty. This development continues until maturity, but with further increase in age this tissue is gradually replaced by fat [14]. But it should be noted that if the older women begins taking hormone replacement therapy (HRT), the proportion of fibroglandular tissue increases at the expense of fatty tissue [15]. It is the fibroglandular tissues within the breast which are believed to have a risk of radiation induced carcinogenesis [14].

![Fig.2.6. Lateral view of an adult female breast. Redrawn from [12] with some modifications](image)

Although the size of the breast varies widely, in the well-developed female it may extend from the clavicle (collarbone) to the sixth or seventh rib (Fig.2.6) and from the lateral border of the sternum to beyond the anterior axillary fold. The average thickness of a compressed breast varies from 45mm to 55mm, depending upon population [14]. For age 50 to 64 years, the average glandularity of a 53mm compressed breast was found to be 29% [16].
2.4.2 Subject Contrast

In X-ray imaging there is always a trade off between subject (radiation) contrast, which is generated by different attenuation of X-radiation by tissue, and SNR. At lower energies, the attenuation coefficients for tissue increase, and this results in better subject contrast. X-ray transmission, on the other hand, is statistically reduced, and this leads to lower SNR. A compromise between subject contrast and transmission results in an optimal energy at which maximal SNR can be obtained at certain dose. Optimal energy for an ideal mammography system is 21 keV for imaging small to medium (4 – 6 cm thickness) and average composition (50% adipose and 50% glandular) breasts. Larger (8 cm and greater) and dense (higher than 50% glandular) breasts require higher energy, which ranges between 16 and 27 keV [13,14].

For breast imaging, subject contrast is also challenging due to the nature of breast tissue. Breast is comprised mainly of adipose and glandular tissues which have very similar attenuation coefficient, certainly at 30 keV and above, as shown in Fig.2.7. This leads to a reduced subject contrast of the breast [10] being imaged. In addition, it is necessary to demonstrate (if any) microcalcifications and malignant tissue in the breast image. Although microcalcifications have much higher effective Z number than that of surrounding tissue and therefore high subject contrast (Fig.2.7), they are usually very small in dimension (about 0.01 mm²) and therefore hard to detect [14]. The malignant tissues are, on the other hand, quite large in size (up to several mm) but, as shown in Fig.2.8, they have a linear attenuation coefficient close to that of the surrounding normal tissue and therefore cause low subject contrast [17].

2.4.3 Mammography X-ray Spectra

The selection of the X-ray spectrum for mammographic imaging is itself a compromise between the needs for high subject contrast and low breast dose. The X-ray spectrum depends on a number of factors. These include anode material, nature and thickness of filtration, and thickness and composition of the breast. The anode material of a mammography system is typically made of molybdenum (Mo). The use of Mo is advantageous because the energy of its main characteristic x-ray lines (Kα and Kβ = 17.5 keV and 19.5 keV, respectively) is near the optimal energy for imaging breasts in average size (4 – 6 cm thickness) and composition (50% adipose and 50% glandular)
glandular). For large and dense breasts, where its characteristic lines have too low energy, the use of Mo would instead enhance patient dose. Therefore, anode materials which are made of other than Mo, such as tungsten (W) and rhodium (Rh), are also used in mammography for imaging less transparent breasts. Rh is chosen because of its main characteristic x-ray lines ($K_a$ and $K_B = 20$ keV and 23.2 keV, respectively) which are greater than those of Mo and therefore, higher mean energy, as shown in Fig.2.9. Tungsten has an advantage that its main $\kappa$-absorption edges ($K_a$ and $K_B = 69.5$ keV and 80 keV, respectively) are beyond mammography energy range and therefore, is more flexible in adjusting mean energy, as demonstrated in Fig.2.10. This makes the W anode material superior for imaging very large breasts [13]. Generally, the use of these anode materials will help reduce the dose to the breast while maintaining the same image quality [19]. Therefore, some manufacturers of full-field digital mammography systems have introduced new models which incorporate dual track anodes with Mo for imaging small to medium breasts and either Rh [20] or W for thick breasts [21].

Fig.2.7. Mass attenuation coefficients of breast tissue and calcifications. Reproduced from [18].
From the tube, an X-ray spectrum emerges through a window of \( \approx 100 \) \( \mu \)m thick beryllium (Be) foil. The combination of the beryllium window with the other X-ray tube components, such as the insulating oil, the glass insert and the target material itself, forms what is called inherent filtration. At this stage, the X-ray spectrum would emerge with a large proportion of lower-energy photons which contributes nothing to the image but instead increases dose to the subject. To solve this problem, an added filter is interposed between the X-ray tube and the subject. This makes the X-ray spectrum harder by shifting it towards higher energy, resulting in a reduced breast entrance surface air kerma (ESAK), which is defined as the kinetic energy released per unit mass in material by ionising radiation, and measured at breast surface (skin). This leads to a reduction in breast mean glandular dose (MGD), i.e. the average dose absorbed by glandular tissue which is presumed to be directly proportional to the risk of carcinogenesis in the breast [22]. ESAK and MGD are the most frequently used indicators of radiation exposure in screening mammography. More details about these two measures and the relationship between them are given later in the following chapters.
Mo anodes are usually combined with 30 μm thick Mo added filters resulting in suppression of photon energies less than 15 keV and a sharp cut-off above the energy at which the K-absorption edge (20 keV) of the filter takes place, as shown in Fig. 2.11. The thickness of Mo filter was chosen as this gives a better compromise between the intensity of the main peak of the characteristic x-rays and that of the
lower energies of the spectrum. Fig. 2.12 shows the X-ray spectra produced by Mo anode combined with 20 μm, 30 μm and 40 μm Mo filters.

![Graph showing X-ray spectra](image)

Fig. 2.11. The effect of using a 0.03 mm Mo added filter on a Mo X-ray spectrum. Reproduced from [6]

In addition to Mo filter, a 25 μm thick Rh filter is also used. Because of its K-absorption edge of 23.2 keV, Rh filter is suitable for all three anode materials mentioned earlier [20, 21, 23]. Materials, such as aluminium (Al), which have no K-absorption edges in this energy range (K-edge of Al is 1.6 keV), were also introduced as an added filter for X-ray tubes with W anode material [24, 25].

### 2.4.4 Breast Compression, Automatic Exposure Control and Focal Spot Size

To further address the problems of variation in the subject contrast in the breast, firm compression is applied using a plastic compression paddle. By doing so, *image contrast*, which is transformed from the subject contrast, by the aid of the image receptor, into differences in optical density in the radiograph or differences in pixel value in the digital image, is also improved. This improvement is due to the reduction in the amount of scattered radiation leaving the bottom of the breast. Compression technique is also employed to lower and uniform breast thickness. While minimising breast thickness results in a reduced radiation dose received by the breast, uniform breast thickness allows the breast to be imaged over a large projected area and
therefore, reduces the overlap of the breast structures in the image. This makes the breast image easier to interpret. Finally, compression paddle is necessary to immobilise the breast in the correct position leading to a reduction in image unsharpness, which is the lateral spreading of the image of a structural boundary. This unsharpness is called *motion unsharpness* which can be further minimised by using a short exposure time. This can be achieved using an automatic exposure control (AEC) which incorporates an ion chamber located under the cassette of a conventional (or the detector of a digital) mammography system. The AEC operates by terminating the exposure when a preset amount of transmitted radiation, based on a preselected film density (pixel value in digital systems) has been detected.

![Figure 2.12](image)

**Fig.2.12.** The Mo X-ray spectra as a function of Mo filter thickness. Using a 0.03mm Mo filter produces a better compromise between the photon flux of the main characteristic X-ray peak and that of lower energies. Reproduced from [6]

However, image unsharpness is also affected by the size of X-ray source (*focal spot*), which is determined by the design of the filament of the cathode; anode diminution and angle; and electron optics that guide the electrons from the cathode to the anode. This is called *geometric unsharpness* which may be reduced by the use of a small focal spot as small as possible. The typical focal spot size used in mammography is 0.3 mm – 0.5 mm.
The use of compression technique may minimise the amount of scattered radiation recorded by the image receptor, but not necessarily eliminate it. Therefore, anti-scatter grid is used. A detail description of the grid can be found elsewhere [14]. Although the use of the grid improves image contrast, by absorbing the scattered radiation before it reaches the film (or detector), it leads to increased radiation dose by 2 – 2.5 times [5], due to the tendency of using higher mAs to compensate the absorbed scattered radiation. In digital mammography systems, however, this effect may be diminished due to the higher efficiency (compared to S/F cassettes) of the detectors used in these systems. Alternatively, a magnification technique, in which an air gap between the receptor (the S/F cassette or detector) and the object being exposed is employed, can be used to reduce the effect of secondary radiation. However, care must be taken that significant reduction in scatter radiation is not achieved with an air gap less than 25 cm [5].

2.5 Image quality

2.5.1 Relative Noise and Image Contrast

In digital imaging, image contrast is defined as the measure of the difference in the mean pixel value between two adjacent locations in an image. In mammography, image contrast is usually measured by exposing a very thin object (e.g. 10mm x 10mm x 0.2mm aluminium foil) placed within a uniform background (e.g. Plexiglas of a range of thicknesses, usually 20mm – 70mm) [26]. Plexiglas was chosen due to its attenuation coefficient which is very close to tissue, as shown in Fig.2.13. Image contrast (C) is then calculated as the difference in the mean pixel value measured within the Al area and that measured in a 2cm x 2cm region of interest (ROI) in the background area, using Eq. 2.2 [26],

\[
C = \frac{\text{mean}(\text{obj}) - \text{mean}(\text{bgd})}{\text{mean}(\text{bgd})}
\]  

(2.2)

where mean (bgd) and mean (obj) are the mean pixel value in the background and in the aluminium square, respectively.
It can be seen from Eq.2.2 that $C$ is not an absolute value, but it is instead normalised to the mean pixel value of the background of an image, which is linear, in direct radiography (DR) systems, or linearised, in computed radiography (CR) systems, to the dose used to acquire the image (a detailed explanation of linearisation process is found in chapter 4). This means that $C$ is independent of dose for a given beam quality and object thickness and therefore, it cannot be used for evaluating digital systems. In addition, it says nothing about the noise which overlays the image information. Here, the noise is defined as the fluctuation in the mean pixel values within the area of a digital image, and thus it is very important parameter of optimising and evaluating digital systems.

In practice, the noise of an image can be calculated as the variation in mean pixel values over a 2cm $\times$ 2cm ROI of the image. In other words, it is the standard deviation of the mean pixel value in the ROI. Generally, the noise in a digital imaging system is assumed to comprise three components; electronic noise, which is not related to the X-ray exposure and is usually called the added noise; structural noise, which is related to the material structure of the X-ray detector; and quantum noise, which depends on the number of photons striking the detector. These three components of noise are in a relationship described [27], as


\[ \sigma_p = \sqrt{k_e^2 + k_q^2 p + k_s^2 p^2} \]  

(2.3)

where \( \sigma_p \) is the standard deviation of the mean pixel value \( p \) in the ROI and \( k_e, k_q \) and \( k_s \) are the coefficients of electronic, quantum and structural noise, respectively.

It was assumed that the image noise is strongly dependent on the dose incident on the detector and therefore on the mean pixel value of the image. In fact, the relationship between noise and mean pixel value of an image was found empirically to be approximated by a simple power relationship described [27] by,

\[ \frac{\text{Noise}}{p} = k_i p^{-n} \]  

(2.4)

where \( \text{Noise}/p \) is known as relative noise, \( k_i \) is a constant and \( n \) equals 0.5 when the noise is purely quantum noise. In practice, however, \( n \) can be slightly lower or higher than 0.5 because of the presence of electronic and structural noise. Clearly, relative noise of an image is a more useful measure than the absolute noise because the relative noise is, as indicated by its name, normalised to the mean pixel value of the image which has a linear relationship with the incident dose used to acquire the image and therefore, can be used for comparing different systems.

2.5.2 Contrast Detail Analysis

The minimum contrast required to detect an object of a certain size at a given dose by a human observer is known as the threshold contrast, and a plot of this threshold as a function of the object diameter is called contrast detail curve. The study to determine the contrast detail curve of an imaging system is known as contrast detail analysis. This analysis has two advantages. The first is that it helps predict the human observer performance of an imaging system for a specific clinical application [28]. The other advantage is that it can be used when different imaging systems are compared at a given dose. This will be explained and discussed in depth in chapter 5 where the contrast detail analysis is used experimentally.

Contrast detail analysis is usually performed using a test object called Contrast Detail (CD) phantom. The phantom consists of Plexiglas (or aluminium) bases with holes (or
metal discs) of different diameter and depth arranged in a matrix of a number of rows and columns. When the phantom is imaged, observers are asked to read the image by specifying the location of each disk in each cell of the matrix. Analysis of the image readings are done using methods described by the manufacturer of the phantom. These methods, however, depend on the specifications of the phantom which vary between radiological applications (e.g. radiography or mammography). A detailed description of the phantom and the methodology of using it in the radiography can be found elsewhere [29].

In mammography it is essential that objects with very small diameter and low contrast can be distinguished from the background. Therefore, to perform the contrast detail analysis for a mammography system, it is necessary to use an appropriate phantom (i.e. different from the phantoms used in general radiography). The most common type used in mammography, and in particular digital systems, is the so-called contrast detail mammography (CDMAM) test object which is described in detail in the next section.

I. CDMAM Phantom

Several phantoms have been designed to assess the image quality of mammography systems. These are not essentially suitable for use with digital mammography systems, in particular those with line patterns. This is because of the fact that dot patterns would represent microcalcifications, which are the smallest objects visible in mammograms, better than lines patterns [30]. Also, phantoms, such as the American College of Radiology (ACR) accreditation phantom, which contain features that simulate breast lesions of interest and used to assess the image in S/F mammography, were found to be unsatisfactory for assessing image quality in digital mammography [31]. The only phantom designed at present with dot patterns is the CDMAM phantom. This was developed as a result of the project: “Quality Assurance in Mammography, at the Department of Diagnostic Radiology, University Medical Centre Nijmegen, St Radboud, The Netherlands” [32]. The current version of the CDMAM phantom (version 3.4), shown in Fig.2.14, consists of a 0.5 mm thick aluminium base, a matrix of square cells and a 5 mm thick Plexiglas cover with overall dimensions of 180 x 240 mm. In each of the 205 cells of the matrix there are two identical gold disks of given thickness and diameter, one is at the centre and the
other is located in a randomly selected corner. The disks vary in thickness (0.03 – 2\(\mu\)m) horizontally and in diameter (0.06 – 2 mm) vertically.

![Image](image.png)

**Fig.2.14.** The CDMAM – phantom (version 3.4). Adapted from [32]

The radiation (subject) contrast of the gold disks is given by,

\[
\text{Radiation contrast} = \frac{I_0 - I_g}{I_0}
\]  

(2.5)

where \(I_0\) and \(I_g\) are the X-ray intensities before and after a gold disk. The linear attenuation coefficient of gold \(\mu\) is given by,

\[
\mu = \frac{1}{x} \cdot \ln\left(\frac{I_g}{I_0}\right)
\]

(2.6)

where \(x\) is the thickness of a gold disk. Substituting Eq. 2.5 by Eq.2.6 gives,

\[
\text{Radiation contrast} = 1 - e^{-\mu x}
\]

(2.7)

For \(\mu\) of gold = 0.190 \(\mu\)m\(^{-1}\) [30], the range of gold disk thickness in the current version of the CDMAM phantom provides an approximate radiation contrast range of 0.56 – 32 %. The contrast range can only be given approximately because the \(\mu\) of gold varies with X-ray energy spectrum and exposure conditions.
II. Human Readings of CDMAM image

When CDMAM image is acquired, it needs to be evaluated using a “Score form CDMAM-phantom” provided with the CDMAM phantom. At least three experienced observers are required for this purpose. The scoring of the CDMAM image is performed by indicating the location of the corner disk of each cell. The indicated location of the disks is then compared to the true disk-locations in the phantom using the “Evaluation form CDMAM-phantom” provided with the phantom. The observer is asked to indicate the disk in at least three cells in each column and each row, in order to apply the nearest neighbour correction (NNC) scheme described below.

In the CDMAM image evaluation, there are three probabilities for each scoring. These include: T if the disk is correctly indicated, F if the disk is incorrectly indicated and N if the disk is absolutely not indicated. When the whole matrix of the image is scored, the NNC scheme is applied [32] as,

1. A True indicated disk requires at least two correctly indicated nearest neighbours to remain True, as shown in Fig.2.15, otherwise it will be regarded False, as shown in Fig 2.15.

2. A False or Not indicated disk needs at least three correctly indicated nearest neighbours to be considered as True. Fig.2.15 illustrates that the False remains False because it has only two correctly indicated nearest neighbours. Fig.2.16 gives an example of a False which is considered as True because it has four correctly indicated nearest neighbours.

3. A True indicated disk which has only two nearest neighbours (i.e. at the edge of the phantom) requires only one correctly indicated nearest neighbour to remain True, as shown in Fig.2.17.

4. A False or Not indicated disk which has only two nearest neighbours will be considered as True when both nearest neighbours are correctly indicated. Fig.2.17 shows an example of a Not indicated disk at the edge of the phantom which is considered to be True because it has two out of two correctly indicated nearest neighbours.

5. The disk in the absent corners of the phantom (0.03 μm/2.0 mm and 2.0 μm/0.06 mm) needs the two nearest neighbours to be correctly indicated to be
considered as True. Fig.2.17 illustrates the two possibilities of the observations of the disk in the absent corners of the phantom.

After applying the NNC rules, a smooth line can be drawn throughout the scoring sheet dividing the cells matrix into two parts. The cells with the true scores will be on one side and those with the false scores will be on the other side, as shown in Fig.2.18. The smallest (threshold) gold thickness for a correctly identified disk is then obtained for each diameter. This can be done by reading the threshold thickness from the gold thickness listed at the side of the CDMAM phantom. The results can be then presented in the so-called contrast detail curve in which the gold thickness is plotted against the corresponding diameter.

In fact, measurement of contrast detail curves by traditional human observation requires readout of multiple phantom images. This approach, however, suffers from two main disadvantages. The first is the low reproducibility and the high variability of the human observer due to the presence of significant inter-observer error. The other disadvantage is that using human readout is tedious work and time consuming. Therefore, a software tool has been developed to perform an automatic scoring of CDMAM images. The software is called CDCOM which was developed by Karssemeijer and Thijssen to perform quick and reliable comparisons of digital mammography systems [30]. This is described in detail in the next section.

III. Automated Readings of CDMAM image

CDCOM software performs fully automatic scoring of digital phantom images in six main steps [4,33]. These are: (1) determination of the grid lines of the phantom in the image matrix. This can be done using Sobel operator and Hough transform techniques. (2) determination of the location of the area in each of the corners of each cell in which a disk could be present, as shown in Fig.2.19a. (3) creating a ROI in each of the four corners of each cell, as illustrated in Fig.2.19b. (4) selecting the corner that is most likely to contain the disk by measuring the highest mean pixel value, as shown in Fig.2.19c. (5) checking whether the right corner is selected by comparing the results with the actual location of the disk of each cell. (6) detecting the centre disk in combination with the three corners of each cell not containing a disk, Fig.2.19d.
Fig. 2.15. Example of the first and second rules of the NNC scheme.

Fig. 2.16. Example of the first and second rules of the NNC scheme.

Fig. 2.17. Example of the third, fourth and fifth rules of the NNC scheme.
Fig.2.18. Score form CDMAM-phantom with the NNC scheme. The smoothing is represented by the thick black line. The arrows indicate the method of obtaining the threshold thickness for the corresponding diameter. For 0.31 mm diameter, the threshold gold thickness is 0.16 μm

The final output of the CDCOM is two separate files, each in the form of a 16 × 16 matrix of numbers. The first file indicates the results for the corner disks and the second file contains the results for the centre disks. As shown in Fig.2.20, each cell of the matrix has a number of 1, 2 or -1, where 1 means a correctly detected disk, 2 is an incorrectly detected disk and -1 indicates a cell that is not in the CDMAM phantom.

In fact, the CDCOM was developed only to analyse the detectability of the gold disks in a single digital CDMAM image. This is not enough for determining threshold contrasts of a digital mammography system. Therefore, further analysis has to be carried out by the user. This includes the combination of the results of multiple CDMAM images and the determination of the threshold contrast, in a method described by Karssemeijer and Thijssen [30] and more recently by Veldkamp et al [34]. The CDCOM manual [33], however, suggests that at least eight images (producing sixteen matrices) should be used for threshold contrast determination.
2.5.3 Modulation Transfer Function

Technically, the "resolution" of a system is defined as the minimum distance that two objects can be placed and still be distinguished by the system as distinct objects. Practically, this meaning of resolution is, however, not very useful, because it depends to some degree on the size and shape of the objects used [28]. However, the situation changes when the transfer of the sinusoidal signals is considered [35]. It is more likely to characterize a sinusoidal distribution imaged by a system in terms of modulation
which is defined as the ratio of the amplitude to the average value of that distribution [36]. The ratio of the output modulation to the input modulation along with the phase shift, when expressed as a function of spatial frequency, is called the optical transfer function (OTF) of the system [36]. It is also defined as the Fourier transform of the two-dimensional point spread function (PSF) or the one-dimensional line spread function (LSF) of the system [37]. The modulation transfer function (MTF) is the absolute value of the complex OTF. In other words, it is the amplitude of the OTF without the phase shift.

![Table](image)

<table>
<thead>
<tr>
<th>Diameter (mm)</th>
<th>Gold thickness (um)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.03</td>
<td>0.04 0.05 0.06 0.08 0.10 0.13 0.16 0.20 0.25 0.36 0.50 0.71 1.00 1.42 2.00</td>
</tr>
<tr>
<td>2.00</td>
<td>-1 1 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>1.60</td>
<td>1 1 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>1.25</td>
<td>1 1 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>1.00</td>
<td>1 1 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>0.80</td>
<td>1 1 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>0.63</td>
<td>2 2 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>0.50</td>
<td>2 2 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>0.40</td>
<td>2 2 1 1 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>0.31</td>
<td>2 2 2 2 1 1 1 1 1 1 1 1 1 1</td>
</tr>
<tr>
<td>0.25</td>
<td>2 2 2 2 2 2 2 2 2 2 2 2 2 2</td>
</tr>
<tr>
<td>0.20</td>
<td>-1 1 2 2 2 2 2 2 2 2 2 2 2 2</td>
</tr>
<tr>
<td>0.16</td>
<td>-1 -1 2 2 2 2 2 2 2 2 2 2 2 2</td>
</tr>
<tr>
<td>0.13</td>
<td>-1 -1 -1 2 2 2 2 2 2 2 2 2 2 2</td>
</tr>
<tr>
<td>0.10</td>
<td>-1 -1 -1 -1 2 2 2 2 2 2 2 2 2 2</td>
</tr>
<tr>
<td>0.08</td>
<td>-1 -1 -1 -1 -1 2 2 2 2 2 2 2 2 2</td>
</tr>
<tr>
<td>0.06</td>
<td>-1 -1 -1 -1 -1 -1 2 2 2 2 2 2 2 2</td>
</tr>
</tbody>
</table>

Fig.2.20. An example of one of the two final output matrices of the CDCOM software. Note, the segmented colours are added for display purpose only. 1: a correctly detected disk, 2: an incorrectly detected disk and -1: a cell that is not in the CDMAM phantom.

Digital imaging systems produce images in which continuous image brightness is represented as a sequence of numerical values. To do so these systems must sample the continuous-image signal at discrete points (samples) to form a discrete-image signal that represent the continuous-image signal. This process is called “sampling”. If the signal is represented by equally spaced samples, or pixels in integrated digital detectors, the space interval between these samples, or pixels, is known as “sample interval,” or “pixel pitch”. The minimum sampling frequency, for complete representation of the continuous-image signal is referred to as Nyquist frequency. Nyquist frequency \( f_n \) of a digital imaging system is given by: \( f_n = f_s / 2 \) where \( f_s \) is the sampling frequency, which is calculated as one over the sampling interval (e.g. pixel pitch) of the digital imaging. Sampling a signal that contains frequencies greater than the Nyquist frequency of the system is referred to as undersampling in which the
system is unable to reconstruct these frequencies, but they would instead overlap with lower frequency components of the signal and appear as image artefacts in a phenomenon known as "aliasing" [38].

When the system is undersampled, as in the case of digital systems, it is more common to use the presampling MTF (preMTF), which describes the response of the system up to, but not including, the sampling of the signal. The preMTF is the result of analogue input subsystems i.e. the response of the system to the blur from an X-ray detection sensor (such as the optical blur in a phosphor layer) and the aperture function of the system. The aperture function may include different things, depending on the design of the system. In a laser-scanned computed radiography (CR) system, the aperture function includes the beam spot response function. In flat-panel imagers (FPI), the aperture function may simply be the shape and size of the active response area of the pixel [28]. The preMTF of a digital system is given as the product of MTF_{sensor} and MTF_{aperture}, where the MTF_{aperture} is given [39] by,

\[
MTF_{aperture} = \frac{\sin(\pi ub) \sin(\pi vb)}{\pi^2 b^2 uv}
\]

where \(b\) is the pixel size, \(u\) and \(v\) are the spatial frequencies in the two dimensions.

In fact, the MTF of the digital system is phase dependent, and hence not spatially invariant [36]. Therefore, the expectation MTF (EMTF) is introduced as a solution to the phase dependence of MTF [37]. It is here defined as the average of the MTF over all phases. However, either EMTF or preMTF can be used to compute the DQE of the detector up to the Nyquist frequency [28, 39], which is one-half of the sampling frequency (rate).

**2.5.4 Noise Power Spectrum**

The noise power spectrum (NPS) of a uniform image is calculated as the well ensemble average (i.e. the mean of a number of ROIs that cover the entire image) of the squares of the absolute value of the Fourier transform of that image and is given by,
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\[ NPS(u, v) = \frac{\langle |FT\{uniform\ image\ (x, y)\}|^2 \rangle}{N_x N_y} \Delta_x \Delta_y \] (2.9)

where \[\langle |FT\{uniform\ image\ (x, y)\}|^2 \rangle\] is the ensemble average of the squares of the value of the 2-D Fourier transformed uniform ROIs, \(\Delta_x\) and \(\Delta_y\) are the pixel sizes in the \(x\) and \(y\) dimensions and \(N_x\) and \(N_y\) are the sizes of the individual square ROI. It has been noted [40] that the size of ROIs used for the estimation of the NPS has to be a matter of great concern. However, it has been found [40,41] that the smallest ROI size that could be used without appreciably biasing the NPS results was 128 x 128 pixels. Other authors [42,43] have found the smallest ROI size to be 256 x 256. NPS is usually divided by the square mean pixel value to give the normalised noise power spectrum (NNPS). NNPS of an image is, as stated for the relative noise in section 2.5.2, a more useful metric than the NPS because the NNPS is relative to the mean pixel value of the image which has a linear relationship with the incident dose used to acquire the image and hence, is useful for comparing systems.

2.5.5 Detection Quantum Efficiency

The ability of a system to transfer the information from the incident X-ray to the output signal is often described in terms of the spatial frequency dependent detection quantum efficiency (DQE \((f)\)) of that system. The DQE has been defined to express the degradation in information signal to noise ratio (SNR) caused by the system relative to the input information and is given by: \(SNR_{output}^2 / SNR_{input}^2\). Since an ideal system extracts all the information in the beam, \(SNR_{ideal}^2 = SNR_{input}^2\), i.e., DQE=1 for ideal system. In general, a totally absorbing photon counting system yields DQE=1. A totally absorbing energy integrating system yields DQE<1 in cases with polyenergetic incident beam [44]. Clearly, DQE is regarded as the best single fundamental measure to describe the performance of digital radiographic systems. Unfortunately DQE is difficult to measure in clinical practice [45]. However, for simplicity the DQE of an imaging system can be estimated as,

\[ DQE(f) = \frac{NEQ(f)}{qK_a} \] (2.10)
where NEQ is the noise equivalent quantum defined as the effective number of quanta used by the detector to give the output signal. In other words, it is the square of the maximum available SNR as a function of spatial frequency given as,

\[ NEQ(f) = [SNR(f)]^2 = \frac{[MTF(f)]^2}{NNPS(f)} \tag{2.11} \]

where MTF is the modulation transfer function (preMTF or EMTF), NNPS is the normalised noise power spectrum. It should be noted that the NNPS is calculated from the NPS which is already squared as it is the variance \( \sigma^2 \) of the pixel to pixel fluctuations in a uniform image of a digital system. \( q \) represents the average number of photons incident on the detector surface per unit area and unit exposure, in air kerma \( (K_a) \), and is given by,

\[ q = K_a \left( \int_0^{E_{max}} \left[ \frac{\Phi(E)}{K_a} \right] dE \right) \tag{2.12} \]

where \( \Phi(E) \) is the normalised spectral air kerma distribution of the incident X-ray beam and \( E_{max} \) is the maximum energy in the spectrum. In fact the DQE is not only a spatial frequency dependant quantity but it is also an exposure dependent quantity and the effect of the photon fluence, per unit area, on the DQE has been well demonstrated elsewhere [46].

References

5. G. T. Barnes and G. Donald Frey, Screen Film Mammography, Imaging Considerations and Medical Physics Responsibilities, Medical Physics Publishing, University Avenue, Madison, Wisconsin 1991


15. S. T. King, Measurement of Film Contrast and Density in Mammograms, an MSc dissertation (1997), University of Surrey


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32. K. R. Bijkerk, M. Thijsen, and Th. Arnoldussen, Manual CDMAM-phantom type 3.4, University Medical Centre Nijmegen St Radboud, Nijmegen, Netherlands, 2002


44. M. Sandborg, Calculation and analysis of DQE for some image detectors in mammography, Report 86, Linköping University, Sweden, 1998

45. K. Bacher, Evaluation of image quality and patient radiation dose in digital radiology, a PhD thesis (2006), Ghent University, Brussels, Belgium

Chapter 3

Literature Review

The digitizing of radiography was successfully performed a long time ago. There are, however, several debates about mammography going digital. A compromise has to be reached between contrast and spatial resolution before deciding on whether to adopt a particular system. A review of the current status of digital mammography and its development will be covered in this chapter.

3.1 Digital vs. Conventional Mammography

Screen-film (S/F) mammography has been studied extensively, and through several large randomized screening trials it is known to reduce breast cancer mortality. Although mammography is currently considered as the gold standard for early detection of breast cancer, conventional S/F imaging is still imperfect. Therefore, the decision to develop digital mammography has been made. Digital mammography has great potential advantages which would effectively overcome the limitations which accompany the analogue technique. Increasing dynamic range, improved quantum efficiency, and storage and display mechanisms are the most significant advantages of the digital systems. Despite these evident advantages, the U.S Food and Drug Administration (FDA) has been careful in approving this new technology [1].

Digital mammography seems to have much wider dynamic range than a conventional system thanks to the digital detector which has an exposure dynamic range wider than the limited dynamic range of the S/F combination [2-4]. An exposure dynamic range at least equal to 100 (two to four times the dynamic range of the typical S/F technique) allows a digital detector to provide a large ratio between the X-ray attenuation of the most radiolucent and most radio-opaque paths through the object (e.g. breast) to be included on the same image. Then, an increased contrast resolution
gives the capability to differentiate structures of interest with lower relative contrast to
the background, compared to film [5]. It should be noted that the dynamic range of
digital mammography is limited by the generator/tube combination not by the
dynamic range of the X-ray receptor as in the conventional mammography. In fact,
the digital detector itself has an estimated dynamic range wider than that of the entire
digital system by a factor of 4 [4].

The superiority in dynamic range as well as the ability to manipulate the image data
after acquisition in a digital system leads to improved sensitivity and specificity [2].
In signal detection theory, sensitivity of a medical imaging system is obtained by
knowing the true-positive fraction (TPF) which can be calculated as: probability of
true positive (TP) / probability of TP + false positive (FP). TP means that the observer
says a signal is present when it is, whereas FP indicates that the observer says the
signal is there when in fact it is not. Specificity is given by (1-FPF), where FPF is the
false-positive fraction = probability of FP / probability of TP + FP.

However, in the most optimal conditions, isolated spherical calcifications with
diameters smaller than 130μm cannot be detected by S/F mammography, even though
its spatial resolution reaches up to 15 mm⁻¹. It has been suggested [6] that the
distinction between malignant and benign microcalcifications and the differentiation
between malignancies themselves can be improved by digitization with a 100 μm
pixel size.

Despite the fact that film performs very well in producing good spatial resolution up
to 15-20 mm⁻¹ [5], the noise associated with such a system would significantly reduce
this advantage. Conversely, digital mammography exists with relatively lower spatial
resolution but with minimum noise thanks to its greatly efficient detector in which the
random fluctuation is approximately limited to X-ray quantum noise. Nevertheless,
the only way to have a good quality image is to have a good enough intensity signal
with very low noise, in other words, a high signal-to-noise ratio (SNR). This depends
on the efficiency of the detector to detect incident photons and create a signal in order
to produce the final image. It can be expressed as detection quantum efficiency (DQE)
[3] which more closely represents the image quality and has been used as a suitable
tool to compare similar S/F systems [5] and to evaluate the performance of digital X-
ray imaging detectors [7]. In general, digital detectors perform with higher DQE than
what would be expected from S/F systems. DQE of a digital mammography with amorophus silicon (a-Si) detector for example is twice that of a conventional S/F system [3]. This would translate to reduction of the dose delivered to the breast tissue if a similar SNR (or image quality) of the two systems is accepted [5].

In fact there is a trade-off between conventional mammography and digital mammography in respect of spatial resolution and SNR, respectively. However, in breast imaging the detectability of low-contrast structures will typically be limited not by spatial resolution but by lack of displayed contrast or by insufficient SNR [5].

Moreover, the presence of the microcalcifications with vermicular (worm-like) shape is one of the strongest clinical indications that these microcalcifications to be more likely related to malignant than benign growth [5]. It happens that these microcalcifications are not the smallest ones, and therefore the spatial resolution requirements may not be so high. This leads to an increase potential of digital mammography over the analogue one.

In addition, digital mammography offers other potential advantages to be used in a screening programme. In S/F mammography the film must act as an image acquisition receptor as well as a storage and display device. With digital imaging systems, on the other hand, acquisition, storage and display are performed separately permitting each of them to be optimized independently. Transferring the data along with the images from site to site and between health centres would be possible as the softcopy workstation is utilized. Three-dimensional (3D) mammography, contrast medium mammography, dual energy imaging techniques and Computer-Aided Diagnosis (CAD), which would hopefully result in improved visibility of lesions [8], can be applied with the possibility to process the digital images with a computer.

Comparing the performance of digital and conventional mammography in a clinical setting has been the subject of several publications [8-11]. In terms of diagnostic accuracy, no statistically significant difference has been detected between full field digital mammography (FFDM) and S/F mammography [8,9] with comparable detection rates [10]. Both techniques have also similar exposure time and patient dose for thinner breast [11]. For thicker breasts, however, digital systems can be operated with shorter exposure time and lower patient dose [11].
3.2 Current Approaches of Digital Mammography

In the last three decades, there have been several attempts to overcome the limitations of using S/F method in mammography applications. Various approaches have been developed in order to digitise a mammogram. Charge Coupled Devices (CCD), Computed Radiography (CR), Direct and Indirect Flat-Panel Imagers (FPI) are utilized in radiology departments at present. They all have the potential to be used for mammographic investigations with some restrictions and this is the subject of the rest of this chapter.

3.2.1 Charge Coupled Devices

The charged coupled device (CCD) is a detector which uses a CsI(Tl) phosphor deposited on a coupling plate consisting of several million independent optical fibres. In such device, the photon interacts first with a phosphor layer producing a visible light photon. This photon is conducted by the fibre optics from the phosphor to a CCD array which converts the light into an electronic signal that is digitised.

A CCD is currently available as a scanning slot digital mammography system with a detector of approximately 1 cm x 24 cm [1]. During exposure, the collimated narrow X-ray beam and detector are moved in synchrony to cover the image field. The great advantage of using a CCD system is the possibility of dose reduction due to the high efficiency of scatter rejection associated with the scanning slot technique. It, however, requires longer total image acquisition time than the FFDM systems described below.

3.2.2 Computed Radiography

Photostimulable phosphor (PSP) based computed radiography (CR) became commercially available in 1983 [12]. Such a system consists of a photostimulable phosphor plate in which the X-ray energy is deposited and stored. This energy is released by a scanning (red) laser beam resulting in a visible (blue) light with intensity proportional to the incident X-ray energy. The released blue light is then captured and converted into a digital signal using an analogue-to-digital converter (ADC). Finally, the residual light energy on the phosphor screen is completely erased by illumination with intense light. Fig.3.1 shows the laser scanning process of the imaging plate of a CR system.
Imaging performance of the CR system, in terms of MTF, DQE and NPS has been investigated and reported in numerous publications [14-20]. The most obvious finding is that the spatial resolution of the CR system was measured to provide a maximum value of 5 - 7 mm\(^{-1}\) whereas it is in the range of 15-20 mm\(^{-1}\) for a S/F system. It should be noted that the resolution of the CR is mainly degraded at the laser reading stage due to the lateral light scatter in the phosphor layer [15,21].

Nonetheless, CR shows potential relative cost-effectiveness and flexibility in comparison with other approaches of digital mammography. This means that CR can provide a smoother transition into digital mammography for many facilities seeking to take the first step into digital territory [22]. Mammography CR systems have recently been introduced in the UK by a number of manufacturers as a step for digitizing medical images in order to install Picture Archiving and Communication Systems (PACS) across the National Health Service (NHS) [18, 23].

### 3.2.3 Energy Integrating Digital Mammography

In this approach, signals from all interacting photons in a detector element (pixel) are normally accumulated (integrated) and then digitised to give a pixel value which is proportional to the incident photon energy. These energy integrating devices can be operated by two methods, according to the mechanism by which the electronic signals are produced through the interactions of X-rays in direct and indirect X-ray conversion detectors. Fig.3.2 shows the detection process of the two types of digital...
detectors. In the indirect approach, the X-rays have to go through two steps before finally converted into signals. The X-ray photons interact first with a scintillation material yielding a light photon. This photon is then detected by an array of photodiodes. Such a method is known as indirect conversion and the best example of this approach is what is called the hydrogenated amorphous silicon (a-Si:H) system. In this way, the light sensitive imaging function of the film in conventional mammography has been replaced by digital light imaging which means “an evolution of screen-film imaging” [24]. The direct conversion method, on the other hand, uses detectors in which the X-rays are converted directly (Fig.3.2(c)), without an intermediate step, into charges which can be recorded using appropriate electronic readout. The current example of this method is the system which uses the amorphous selenium (a-Se) detector.

I. Amorphous Silicon Flat Panel Imager

For more than a decade, digital X-ray imagers based upon a-Si flat panel array imager (FPI), with a large area, have been under development as promising candidates for application in diagnostic radiology. The form which has received the greatest attention in imaging devices is the hydrogenated amorphous silicon (a-Si:H), i.e. amorphous silicon is permeated with hydrogen and diffused with p and n dopants to provide device junction. The a-Si:H is very sensitive to visible light, with an efficiency close to 1 and a low dark current resulting in a low pixel noise [25]. An example of this type has been produced by General Electric (GE) Healthcare (Milwaukee, WI) [26]. As illustrated in Fig.3.3, a-Si FPI consists of a layer of material, --scintillator screen-- by which the incident X-rays are absorbed and converted into light and a thin film transistor (TFT) deposited in the term of “active matrix” on a large glass to cover a field size of about 19 cm x 23 cm [27]. Several X-ray convertor materials can be used with TFT flat panels, such as Gd2O2S: Tb phosphor screen or thallium doped caesium iodide (CsI:Tl) scintillating material. The array is divided into small elements (pixels) each contains a TFT which acts as an electronic switch. Each TFT element comprises an a-Si photodiode, which converts the light into electron-hole (e-h) pairs, with its associate field effect transistor (FET).
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Fig. 3.2. (a) Phosphor screen indirect, (b) CsI (TI) scintillating layer indirect, and (c) photoconductor direct methods of X-rays detections.

Each photodiode consists of a base metal contact, a p-doped, an intrinsic, an n-doped a-Si:H layer, and a top transparent metal contact. When an adequate reverse bias is applied to the photodiode through the metal, p-n junction becomes fully depleted and the e-h pairs generated in this junction are drifted toward the contacts under influence of the intrinsic and applied electric fields. During imaging, with the FETs non-conducting, the e-h pairs formed in the depletion layer results in a discharging of the sensor capacitance. The degree of this discharging constitutes the integrated imaging information. At the readout stage, one row of pixel capacitors is turned on by activating the FETs and the image charge from those pixels is transferred on to the data lines in order to be amplified using external charge preamplifiers and then to ADC. After a given row is read out, the associated FETs are turned off, and the process continues.
The use of phosphor materials with a relatively high atomic number makes the photoelectric absorption to be the dominant type of X-ray interaction. This leads potentially to the excitation of many electrons in the phosphor and thereby the production of many light quanta. After their formation, the light quanta needs to successfully escape the phosphor and be effectively coupled to the next stage for conversion to an e-h pair and read out. While travelling within the phosphor, the light can, however, spread to adjacent pixels of the diode (Fig.3.2 (a)) resulting in reducing spatial resolution of the imager. A possible solution to this problem is the use of a structured scintillator that consists of small column-like CsI (Tl) crystals (Fig.3.2 (b)). Such a technology avoids lateral diffusion of the emitted light photons by guiding them through very thin needles. As a result, even thick scintillator layers can be considered, resulting in a high DQE and still a high spatial resolution [28].

Fig.3.3. Structure of the CsI(Tl)-α-Si photodiode flat panel imager. Redrawn from [1]

The performance of the flat-panel α-Si has been extensively studied for general radiography [29] and mammography applications [26, 30-33]. The results have shown that the α-Si has a higher performance in comparison with conventional S/F imaging
combinations with respect to low-contrast sensitivity, small-detail visibility, and patient dose [26]. In addition, \( \alpha\)-Si system has also exhibited a much larger DQE and a wider dynamic range than those of S/F and CR systems under mammographic conditions [32]. Although a conventional S/F system has superior spatial resolution over the \( \alpha\)-Si detector, it has been suggested [34] that lesion detectability can be improved even with lower spatial resolution by contrast enhancement of digital data. Furthermore, microcalcification detectability was found [35] to be as good as with \( \alpha\)-Si imager, a S/F system and a CR technique.

However, the selection of X-ray converter, the trade-off between efficiency and resolution (thick or thin scintillator, respectively) and the higher cost of the flat-panel \( \alpha\)-Si are the great challenges needing to be faced before conventional mammography can be entirely replaced with such a system. Finally, the time response behaviour of the detector is crucial for practical considerations. Single images needs not to show ghost images of previous investigations. Real time of the fluorescence process must, therefore, be free from image lag, which is mainly caused by deep trapping in the phosphor layer and emission of electrons from traps in the photodiode, in particularly when high dose images were taken shortly before [36].

II. Amorphous Selenium Flat Panel Imager

The components of the \( \alpha\)-Se FPI active matrix are similar to those of the indirect \( \alpha\)-Si FPI. It consists of a TFT for image readout, a collecting pixel electrode and a storage capacitor to hold the image charge temporarily. The \( \alpha\)-Se FPI differs from the \( \alpha\)-Si FPI in that it does not employ a phosphor layer. Therefore, the X-rays are absorbed by the \( \alpha\)-Se detector and then directly converted into \( e-h \) pairs. An example of this type has been produced by IMS Giotto (Bologna, Italy), with a pixel size of 85 \( \mu \)m and a field size of 18 cm \( \times \) 24 cm [37].

Selenium is an ideal material for a mammography detector, because it has high X-ray absorption efficiency, extremely high intrinsic resolution and low noise [24]. Using this detector as the photoconductor, a thickness of 250 \( \mu \)m is adequate to stop more than 95\% of X-rays in the mammography energy range. Standard screens (e.g. \( \text{Gd}_2\text{O}_2\text{S} \) ) that are used in S/F mammography have only about 50 to 70 \% quantum efficiency, and the scintillator CsI (TI) used in \( \alpha\)-Si technology exhibits about 50 to 80\% quantum efficiency [24].
A comparison between direct and indirect conversion FPIs in terms of MTF, NPS and DQE has been studied and reported elsewhere [38-43]. The most significant findings are that the direct systems exhibit higher MTF with no dose dependence. Indirect systems, on the other hand, impart lower NPS and additive noise than those of direct systems. The explanation is that the loss of light photons due to the pixel fill factor (which equals about 75% of the TFT) does not usually result in a complete loss of X-rays [39], due to the blurring caused by phosphor screens. Therefore, indirect systems have greater DQE than, or at least comparable to, that of the direct one [40] but this is only true at the lower frequencies (1 – 2 mm^{-1}) [41]. At higher frequencies, a direct FPI seems to have better DQE than that of an indirect system. Consequently, it has been concluded [41] that the direct system is preferable when fine anatomic structures need to be imaged with high detail and contrast whereas the indirect is desirable in radiographic applications where the visibility of low contrast anatomic structures is limited by noise.

It is of note that, with unknown reasons, the DQE of the a-Se detector was noticed to decrease rapidly with radiation energy [42]. Also, due to the relative lower atomic number of Se material (Z=34), its photoelectric cross section decreases more rapidly as beam energy increases. Therefore, the way of the users to increase X-ray energy slightly when switching from the S/F to the CR systems is no longer valid when switching to a-Se system [44].

III. Pixelated Compound Semiconductor Detectors

Recently, the development of position sensitive solid-state detectors constructed from wide bandgap compound semiconductors such as cadmium telluride (CdTe), cadmium zinc telluride (CZT), gallium arsenide (GaAs) and lead iodide (PbI₂), which showed rather optimistic results, are being carried out for application to digital mammography [45]. They have high atomic number and high density, which make it possible to construct a very thin X-ray detector that still has high quantum absorption efficiency [46]. A major challenge which faces this approach is to obtain coupling between all of the detector pixels (50 µm x 50 µm each) and the readout electronics. A potential solution for this difficulty is to use the so-called indium bump bonding technique [48]. Fig.3.4 illustrates the concept of a direct conversion detector indium bump bonded to a readout electronic chip. This technique has an advantage over the
TFT method (used in the $\alpha$-Si and $\alpha$-Se detectors) of having a 100% fill factor of each pixel, whereas it is about 75% fill factor in the latter.

Progress has been evaluated of hybrid solid-state pixel detectors for digital mammography [47-49]. Although a single semiconductor silicon (Si) detector has the advantage of easy processing, higher MTF can be obtained with a CZT detector. 25% and 10% MTFs were reported [47,48] at 10 mm$^{-1}$ Nyquist frequency for CZT and Si detectors, respectively. DQE, on the other hand, was better with Si detector than those with CZT and CdTe detectors. This is mainly because of the charge collection inefficiency of CZT and CdTe detectors. While the low hole mobility was the main problem of the CZT detector, the CdTe detector was mainly limited by the polarization effect [47]. In fact, the hole mobility life-time product of the CZT detector is in the order of $10^{-5}$ cm$^2$/V. This means that the total distance that holes can travel under a bias voltage of 300 V/mm is about 0.3 mm and hence, the charges generated by the X-rays will not be fully collected in a detector thicker than this [48]. However, the CZT detector has potential for better performance by using an electron-collecting readout technique that would collect electrons instead of holes.

![Diagram of CZT-PIN Detector Array](image)

Fig.3.4. The concept of a solid-state direct conversion CZT pixelated detector indium bump bonded to a readout electronic chip. From [48].

The main challenge faced in the use of GaAs detector is the small depletion region and the high charge trapping in the present GaAs material [47]. However, optimizing the dopant type and concentration of GaAs material is being carried out [49], which
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may resolve the problem of this detector. PbI₂, in contrast, is a very new detector material and needs further development before it can be used as a pixelated detector [47].

3.2.4 Photon Counting Digital Mammography

Unlike the energy integrating systems, the photon counting digital mammography processes the signals from each single interacting X-ray photon individually. The advantage of this approach over the energy integrating devices is to provide a signal with much smaller detector and electronic noise. This can be done using the so-called noise rejection technique by which the photon counting device rejects the number of pulses below an electronic threshold level. The greatest challenge for photon counting devices is to be able to accommodate the high rate of X-rays interacting per unit time in each pixel of the detector. The possible solution to this problem is to utilise modern electronics and to apply a signal weighting scheme [1].

In the photon counting full field digital mammography a Si detector line and a single beam are operated and moved together during the exposure to fully cover the image field. Such a design is called scanned-slot X-ray imaging. However, to acquire an image in a reasonably short time, multiple beams each matched with a Si detector slit are scanned across the image filed. This geometry is referred to as multi-slit X-ray imaging. Fig.3.5 demonstrates the two different geometries of slit scanning. An example of this type has been recently launched by Sectra Mamea AB (Stockholm, Sweden), with a pixel spacing of 50 μm and a field size of 24.3 cm x 26.5 cm [50].

It was noted that the dose efficiency of a mammography system is highly dependent on the rejection of scattered radiation [51]. Detecting the scattered radiation leads to a decrease in the image contrast which has to be compensated for by increasing number of quanta incident on the detector. This in turn leads to an increase in the dose absorbed by the breast. It was found that using multi-slit scan geometry achieves scatter-to-primary ratio that is lower than any other type of scatter rejection scheme currently employed in mammography systems [52]. This is attributed to the smaller detector area and the low angular acceptance of the post-collimator slits [52] as shown in Fig.3.5. Moreover, the photon counting multi-slit scanning system was found to have a good MTF and DQE over a wide dose range compared to CR and flat-panel systems [20].
3.3 Other breast imaging approaches

As mentioned earlier, the sensitivity of mammography can be significantly affected by fibroglandular tissue in the breast. This means that the sensitivity of mammography is poor for women who have dense breasts, in particular those with an extremely dense pattern on mammograms. This limitation of mammography is compounded by the fact that dense breast tissue is a significant risk factor for developing breast cancer [53].

As discussed earlier, digital mammography has been introduced in an effort to overcome the known limitations of mammography. It has been shown that digital mammography provides a significant increase in sensitivity for the detection of breast cancer. However, the benefits of digital mammography are debatable because its significant benefit was shown in only subgroup of women but not in the general screening population and because of its questionable cost-effectiveness at this time [53]. For these reasons considerable attention has been given to other breast imaging techniques such as sonography, magnetic resonance imaging (MRI) and nuclear medicine (NM). The advantages and drawbacks of these technologies in breast imaging are discussed briefly in the following text.

Currently, sonography provides the greatest support to film mammography. As equipment improves, the use of sonography in breast imaging has become more important. This improvement enables the user not only to detect malignancies that previously were not detectible sonographically, but also to identify benign lesions [54]. In addition, sonography is performed to help determine the need for a biopsy or
additional testing. In spite of all these benefits, sonography is currently not recommended for breast cancer screening in the general population [55]. This is presumably based on the findings of the American College of Radiology Imaging Network (ACRIN) 6666 Trial which compared breast sonography and mammography. The preliminary results from that trial showed that the addition of screening sonography resulted in only a modest increase in cancer detected. Moreover, the results showed the positive predictive value of sonography-prompted biopsies was only 8.5% compared with 29% for mammography [53]. In general, the potentially considerable contribution of sonography is substantially limited by suboptimal imaging technique and inconsistent operator training [54].

In the literature, both mammography and sonography were found to have a low sensitivity for the detection of breast cancer in women with radiographically dense breasts. For MRI, on the other hand, a sensitivity of between 75% and 100% was reported [56-58]. However, although MRI is likely to have an increasing impact on solving mammography and sonography problems as time goes on, the use of such technology is greatly hindered by the expense and time required for breast MRI examinations (10–15 times that of mammography), and the large variation in specificity (29–98%), depending on technique and patient selection [59, 60]. This can possibly explain why breast MRI is still not recommended as a screening exam for the general population of women. But it might be used on women who may need additional imaging after standard imaging exams (mammography and ultrasound), and it may also be helpful in patients with a strong family history of breast cancer [55].

For more than a decade, nuclear medicine has been introduced as an alternative method for breast imaging. This technology uses the radiotracer technetium-99m (99mTc) sestamibi in a technique called "scintimammography" [53]. Because this technique is independent of breast density, it was thought to be particularly useful in imaging women with dense and fatty breasts [61]. Such a technique is, however, limited by the very low spatial resolution of conventional gamma cameras. These cameras suffer also from breast positioning limitations. As a result scintimammography technique had poor sensitivity (< 50%) for the detection of small breast cancers (≤ 10 mm in diameter) [62]. It is interested to note that up to one third
of breast cancers detected on screening mammography and half of those detected on MRI in screening studies are smaller than 10 mm [63,64]. This makes the limitation of scintimammography technique is more important and explain the limited diagnostic value of the technique [53].

However, development in the detectors of both single-photon emission computed tomography (SPECT) and positron emission nuclear tomography (PET), over the past decade, have brought in various designs of small-field-of-view detectors dedicated for breast imaging. The dedicated technologies offer significant improvements in both spatial and energy resolution and allow the breast to be positioned directly on the detector, permitting better detection of small breast tumours [53].

References

1. E. D. Pisano, M. J. Yaffe, and C. M. Kuzmiak, Digital Mammography, Lippincott Williams & Wilking, USA, 2004
9. C. M. Kuzmiak, Comparison of Full-Field Digital Mammography to Screen-Film Mammography with respect to Diagnosis Accuracy of Lesion Characterization in Breast Tissue Biopsy Specimens, Academic Radiology, Vol. 9, No. 12, 2002, Pages: 1378-1382


25. D. Darambarra, Course notes on radiation imaging detectors: MSc in Medical Physics, University of Surrey, 2004


CHAPTER 3. CURRENT APPROACH IN MAMMOGRAPHY

29. X. Liu and C. C. Shaw, a-Si:H/CsI(Tl) flat-panel versus computed radiography for chest
imaging applications: image quality metrics measurement, Med. Phys., Vol. 31, No. 1,
2004, Pages: 98-110

30. D. G. Darambara, A. Taibi and R. D. Speller, Image-quality performance of an a-Si:H-
based X-ray imaging system for digital mammography, Nuclear Instruments and Methods

31. D. G. Darambara, R. D. Speller, J. A. Horrocks, S. Godber, R. Wilson and A. Hanby,
Preliminary evaluation of a prototype stereoscopic a-Si:H-based X-ray, Nuclear

prototype flat-panel imager operated under mammographic conditions, Med. Phys., Vol.
30, No. 7, 2003, Pages: 1874-1890

33. C. Ghetti, A. Borrini, O. Ortenzia, R. Rossi, P. L. Ordóñez, “Physical characteristics of
GE Senographe Essential and DS digital mammography detectors,” Med. Phys. 35, (2),

34. S. Vedantham, A. Karellas, S. Suryanarayanan, D. Albagli, S. Han, E. J. Tkaczyk, C. E.
Landberg, B. Opsahl-Ong, P. R. Granfors, I. Levis, C. J. D’Orsi, R. E. Hendrick, Full
breast mammography with an amorphous silicon-based flat panel detector: Physical
characteristics of a clinical prototype, Med. Phys., Vol. 27, No. 3, 2000, Pages: 558-567

Dryden, T. W. Stephens, S. K. Thompson, K. T. Krugh and Chao-Jen Lai,
Microcalcification detectability for four mammographic detectors: Flat-panel, CCD, CR,

International School on Condensed Matter Physics (ISCMP): Future Directions in Thin
Kirov, A. Vavrek, J. M. Maud (1996) 112-120

37. J. M. Oduko, C. Young, A. Al Sager and O. Gundogdu, Technical Evaluation of The IMS
Giotto Full Field Digital Mammography System with a Tungsten Tube. NHS Breast
Screening Programmes, 2008 (NHSBSP Equipment Report 0804)

38. R. Fahrig, J. A. Rowlands, M. J. Yaffe, X-ray with amorphous selenium: Detective
quantum efficiency of photoconductive receptors for digital mammography, Med. Phys.,

39. W. Zhao and J. A. Rowlands, Digital radiology using active matrix readout of amorphous
12, 1997, Pages: 1819-1833

40. G. Borasi, A. Nitrosi, P. Ferrari, D. Tassoni, On site evaluation of three flat panel

41. E. Samei and M. J. Flynn, An experimental comparison of detector performance for direct
and indirect digital radiography systems, Med. Phys., Vol. 30, No. 4, 2003, Pages: 608-
622

42. Personal conversation with Professor Kenneth C. Young, head of the National
Coordinating Centre for the Physics of Mammography (NCCPM), Medical Physics, St
Lukes Wing, Royal Surrey County Hospital, Guildford, UK

CHAPTER 3. CURRENT APPROACH IN MAMMOGRAPHY


46. T. Schulman, Si, CdTe and CdZnTe radiation detectors for imaging applications, PhD thesis, University of Helsinki, Helsinki, Finland, 2006


Chapter 4

Optimisation of CNR Measurement

4.1 Introduction

The European Guidelines for quality control in digital mammography define a procedure for measuring contrast-to-noise ratio (CNR) using a 0.2 mm thickness of aluminium with different thicknesses of Plexiglas (PMMA) [1]. The method specifies the use of a 2 × 2 cm region of interest (ROI). Some problems have become apparent in applying the guidelines to images from computed radiography (CR) systems that do not arise when testing direct digital radiography (DR) systems. These include:

- the need to correct for the non-linear nature of CR images
- the distorting influence of the uncorrected heel effect
- the effect of the ROI size

One way to address these problems would be to linearise the images, and apply a heel effect correction before calculating the CNR in the specified manner. However, this approach has disadvantages. The first is that two images are required for each CNR measurement which is time consuming. The other is that the heel effect correction procedure adds a substantial complication to what should be a simple measurement.

The aim of this chapter is

a) To investigate the methods of linearising image data

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b) To investigate the consequence of applying or not applying a heel effect correction when measuring CNR with CR systems in comparison with DR systems

c) To investigate the impact of using different sizes of ROI in determining CNR for CR systems using heel effect corrected and uncorrected images

4.2 Materials and Methods

4.2.1 Systems Tested and Image Acquisitions

The study was carried out using three CR systems and one DR system. The DR system was a Hologic Selenia with an amorphous selenium (α-Se) detector. The CR systems used are described in Table 4.1. Images of a 45 mm thickness of PMMA and a 10 mm x 10 mm x 0.2 mm square of aluminium were acquired at different levels of exposure for each system including that selected automatically by the automatic exposure control (AEC) of the X-ray set. An image of the PMMA phantom without the aluminium square (uniformity image) was also acquired for the CR systems to enable heel effect correction. All images were saved as unprocessed DICOM files for later analysis. The detector entrance air kerma was measured using an MDH 2025 electrometer and a 20 x 5-M mammography ionisation chamber (RadCal Corp., Monrovia, CA).

Table 4.1. CR systems used in this work

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>Image plate</th>
<th>CR reader</th>
<th>Pixel size (μm)</th>
<th>Mammography X-ray set</th>
<th>Factors selected by AEC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Agfa</td>
<td>MM 3.0</td>
<td>CR85-X</td>
<td>50</td>
<td>GE Senographe</td>
<td>30kV Mo/Rh 96 mAs</td>
</tr>
<tr>
<td>Kodak</td>
<td>HER-M2</td>
<td>DirectView CR 850</td>
<td>50</td>
<td>GE Senographe DMR+</td>
<td>27kV Mo/Rh 111 mAs</td>
</tr>
<tr>
<td>Konica</td>
<td>RP-6M</td>
<td>Regius190</td>
<td>43.75</td>
<td>GE Senographe DMR</td>
<td>28kVMo/Mo 76 mAs</td>
</tr>
<tr>
<td>Minolta</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

4.2.2 System Response and Linearisation Method

For each exposure level, the mean pixel value and the standard deviation was calculated using a 10 x 10 mm² region of interest (ROI) positioned on the mid-line and 60 mm from the chest wall edge of each image. The characteristic response curve was then obtained by plotting the average pixel values as a function of detector
entrance air kerma, for the DR and CR systems as described in the UK protocol [2]. For the DR system, the response was linear, with a pixel value offset [3]. The offset was found to be 53.3 which was then subtracted before determining relative noise in the images. Fig. 4.1 shows the detector response of the DR system used in this work. The images from the CR systems had as expected non-linear responses, presumably due to pre-processing [4,5]. Fig. 4.2 shows the non-linear response of pixel values to incident air kerma for the images from the Agfa CR system with a power function fit. This relationship was subsequently used to linearise the pixel values of the CR images from this system as shown in Fig. 4.3.

Fig. 4.1. The detector response of the Hologic DR system. The error bars indicate the 2 standard error of the mean of the pixel values.

For the Agfa CR system, it was easy to linearise the images as each image had a simple power relationship between the pixel values and the incident air kerma [6] (Note that it would also have been possible with the Agfa CR system to save the images as a logarithmic relationship). For the other CR systems, however, the relationship was more complex and linearising the images would have been complicated [7] and therefore, the measured mean pixel values and standard deviations from ROI measurements were linearised instead. The procedure for linearising the ROI measurements is described [2] as;
CHAPTER 4. RELATIVE NOISE AND CNR MEASUREMENTS

\[ P = a - b \cdot \log K \]  \hfill (4.1)

Fig. 4.2. Agfa CR response; measured mean value plotted against incident air kerma at the cassette surface. The error bars indicate the 2 standard error of the mean of the pixel values.

Fig. 4.3. Linearised mean pixel value plotted against incident air kerma on the cassette of the Agfa CR system. The error bars indicate the 2 standard error of the mean of the pixel values.
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where, \( P \) is the measured pixel value corresponding to an entrance air kerma of \( K \), and \( a, b \) are fitted coefficients. This can be then inverted to give,

\[
P' = 10^\left( \frac{P-a}{-b} \right)
\]

where, \( P' \) is the linearised value of \( P \). The standard deviations \( \sigma' \) for the linearised pixel values \( P' \) can be obtained as,

\[
\sigma' = 10^\left( \frac{(P+\sigma_p)-a}{-b} \right) - P'
\]

where \( \sigma_p \) is the standard deviation in the pixel values in a \( 1 \times 1 \) cm ROI with an average pixel value \( P \). Both methods of linearisation (i.e. linearising images and linearising data from non-linear images) were used with the Agfa system to confirm equivalence.

4.2.3 Heel Effect Correction

Heel effect correction was achieved by acquiring an image of the PMMA block without the aluminium square (uniformity image). Fig. 4.4 shows a profile along the mid-line parallel to the anode-cathode axis on such an image demonstrating the presence of low and high frequencies noise. The image was then smoothed 20 times using a \( 7 \times 7 \) Gaussian filter to produce an image which retained the low frequency noise (mainly due to heel effect) but eliminated the high frequency noise arising from the detector. Fig.4.5 shows the profile given in Fig. 4.4 after this smoothing process. The smoothed image (Image B in Eq. 4.4) was then used to remove the low frequency noise due to the heel effect in the image with the aluminium square (Image A in Eq. 4.4). This was done by dividing image A by image B and multiplying by a scaling factor to produce image C, as shown in Eq. 4.4. The scaling factor was adjusted to produce an image with average pixel values in the region of the aluminium square that were similar to those of the original image. Fig. 4.6 shows the profile measured in image C (the heel effect corrected image) showing the high frequency noise only.

\[
\text{image } C = \frac{\text{image } A}{\text{image } B} \times \text{scale factor }
\]
Fig. 4.4. A profile measured along the mid-line parallel to the anode-cathode axis on a uniformity image (block of PMMA without the aluminium square), acquired using the exposure factors selected by the AEC of the Agfa CR system. High and low frequencies noise are present.

Fig. 4.5 A profile measured along a smoothed uniformity image after removal of the high frequency noise. (Profile at same location as that shown in Fig. 4.4)
4.2.4 Relative Noise Measurement

The linearised mean pixel value and its standard deviation was determined in the background area of both heel effect corrected and uncorrected images, using a range of ROI sizes (0.25 x 0.25 cm – 5 x 5 cm). To produce ROI with a size of 0.25 cm x 0.25 cm, a ROI was first drawn with an area of 0.5 cm x 0.5 cm and then subdivided into four 0.25 x 0.25 cm sub-ROIs. The relative noise of the CR and DR images was calculated (using Eq. 4.5) for each ROI size. This was repeated for the CR systems after applying the heel effect correction.

\[
\text{Relative noise} = \frac{sd(bgd)}{\text{mean}(bgd)}
\]  

(4.5)

where, \(\text{mean}(bgd)\) is the mean pixel value measured in the background ROI, and \(sd(bgd)\) is its standard deviation.

4.2.5 CNR Measurement

The linearised mean pixel values and their standard deviations were determined in the area of the aluminum square and the background, for both heel effect corrected and
CHAPTER 4. RELATIVE NOISE AND CNR MEASUREMENTS

uncorrected images, using different ROI sizes. For each ROI size, four ROIs were selected around the aluminium square and averaged to give the mean pixel value and standard deviation of the background. Four background ROIs surrounding the aluminium square were used in order to obtain an accurate measure of the mean background level where the background was not uniform due to the heel effect. The number of ROIs and their sizes used in the CNR measurements are shown in Table 4.2. To produce ROIs with a size of 0.25 \times 0.25 \text{ cm}, the four background ROIs and the ROI in the aluminium square were first drawn with an area of 0.5 \times 0.5 \text{ cm} and then subdivided into four 0.25 \times 0.25 \text{ cm} sub-ROIs, as shown in Fig. 4.7. This gave 16 background sub-ROIs and four aluminium sub-ROIs. The CNR of the CR and DR images was calculated (using Eq. 4.6) for each background ROI size. This was repeated for the CR systems after applying the heel effect correction.

Table 4.2. Number and size of ROIs used in the CNR measurements

<table>
<thead>
<tr>
<th>Measurement set up</th>
<th>Size of background ROI (cm)</th>
<th>Number of background ROI</th>
<th>Size of aluminium ROI (cm)</th>
<th>Number of aluminium ROI</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.25 \times 0.25</td>
<td>16</td>
<td>0.25 \times 0.25</td>
<td>4</td>
</tr>
<tr>
<td>2</td>
<td>0.50 \times 0.50</td>
<td>4</td>
<td>0.50 \times 0.50</td>
<td>1</td>
</tr>
<tr>
<td>3</td>
<td>1.00 \times 1.00</td>
<td>4</td>
<td>0.50 \times 0.50</td>
<td>1</td>
</tr>
<tr>
<td>4</td>
<td>2.00 \times 2.00</td>
<td>4</td>
<td>0.50 \times 0.50</td>
<td>1</td>
</tr>
</tbody>
</table>

Fig. 4.7 ROIs used when the measurement area was 0.25 \times 0.25 \text{ cm}.

\[
CNR = \frac{\text{mean(bgd)} - \text{mean(Al)}}{\sqrt{\frac{(sd(bgd))^2 + (sd(Al))^2}{2}}}
\]  

(4.6)
CHAPTER 4. RELATIVE NOISE AND CNR MEASUREMENTS

where, \( \text{mean (} Al) \) is the mean pixel value measured in the aluminum square and \( sd (} Al) \) is its standard deviation, \( \text{mean (bgd)} \) is the average of the mean pixel values measured in the ROIs around the aluminum square, and \( sd (bgd) \) is the average standard deviation of the ROIs around the square.

The European and the UK protocols [1,2] specify a procedure for measuring the CNR using only a single ROI inside the aluminium square and a single ROI outside. The ROI outside the aluminium square (background ROI) is located on one side of the aluminium square, in the direction perpendicular to the anode – cathode axis. This location is chosen to minimize the impact of the heel effect associated with the CR images. The impact of the heel effect and the ROI size on the relative noise and CNR measurement, using this procedure, is also investigated in this chapter.

4.3 Results

4.3.1 Image and Data Linearisation

For the Agfa CR system two methods of correcting for the non-linear nature of the images were used and the results compared in Fig. 4.8. This shows the relative noise of the Agfa CR images plotted as a function of incident air kerma. It can be seen from Fig. 4.8 that the two methods of linearization gave similar results for the mean relative noise, and that this was quite different from that found using simple measurements on the non-linearised images. For the other two CR systems the images themselves were not linearised. Instead, the mean pixel values and standard deviations were corrected using the measured relationship between pixel values and entrance air kerma.

4.3.2 Impact of ROI Size and Heel Effect on Relative Noise Measurement

The relative noise measurements for the Agfa CR images (with and without heel effect correction) are plotted as a function of ROI size in Fig. 4.9 at the AEC selected dose level. Also shown in this figure are the corresponding measurements for the DR system. Figs. 4.10 and 4.11 show the effect of ROI size and heel effect on the relative noise measurements for the Kodak and Konica CR images.
Fig. 4.8 Relative noise plotted as a function of incident air kerma for the Agfa CR system using (a) mean pixel values from non-linear images, (b) linearised mean pixel values from non-linear images and (c) mean pixel values from linearised images.

Fig. 4.9. Relative noise plotted as a function of ROI size for the Agfa CR at the AEC selected exposure parameters, with and without heel effect correction. Relative noise of the DR system, calculated at its AEC selected exposure parameters (31 kVp Mo/Rh, 130 mAs) is included for comparison.
Fig. 4.10 Relative noise plotted as a function of ROI size for the Kodak CR at the AEC selected exposure parameters, with and without heel effect correction.

Fig. 4.11 Relative noise plotted as a function of ROI size for the Konica CR at the AEC selected exposure parameters, with and without heel effect correction.
Fig. 4.12 shows the relative noise measured as a function of ROI size using the procedure proposed by the European and the UK protocols using a single ROI for the background and aluminium square. The results obtained using the method proposed here (i.e. four ROIs around the aluminium square) before and after heel effect correction are also shown for comparison.

![Relative noise graph](image)

**4.3.3 Impact of ROI Size and Heel Effect on CNR Measurement**

The CNR for the Agfa CR images is plotted as a function of incident air kerma on the cassette for different sizes of ROI as shown in Fig. 4.13. Figs. 4.14 and 4.15 show the effect of ROI size on the determination of CNR for the Kodak and Konica CR images. The CNR values calculated in a 2x2 cm ROI using heel effect corrected images are included for all systems.
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Fig. 4.13. CNR plotted as a function of incident air kerma using different ROI sizes for the Agfa CR system. The CNR measurements using 2x2 cm ROIs with heel effect corrected images are also shown.

Fig. 4.14. CNR plotted as a function of incident air kerma using different ROI sizes for the Kodak CR system. The CNR measurements using 2x2 cm ROIs with heel effect corrected images are also shown.
### 4.4 Discussion

#### 4.4.1 Image and Data Linearisation

The relationship between mean pixel values and the incident air kerma were determined for each system. This was linear for the DR system, with a pixel value offset. For the CR systems, the relationship was non-linear as expected. The UK protocol [2] states that the data used for CNR measurement for CR systems should be linearised with respect to the air kerma incident on the detector. This can be achieved by linearising the image or linearising the mean pixel value and standard deviation measured in the image. In this work, both methods were used with the images from the Agfa CR system with essentially similar results. For the other CR systems, the simpler method of linearising the measured data was employed. Where no method of linearising the images or data was employed very different results were found.

#### 4.4.2 Impact of ROI Size and Heel Effect on Relative Noise Measurement

The relative noise was calculated for images from the DR and CR systems using a range of ROI sizes. The dependence of the relative noise measurement on ROI size was substantial for the CR systems. This is attributed to the presence of the heel effect.
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effect. For the DR system, the results showed very little dependence of the relative noise on ROI size as heel effect correction is applied by the manufacturer. In this work, heel effect correction was applied to the CR images used for the relative noise measurement. The relative noise measured in these images showed insignificant dependence on ROI size, for all CR systems tested. The use of small ROI (i.e. 0.25 x 0.25 cm) resulted in relative noise measurements that were very close to those found after heel effect correction.

The European and the UK protocols specify a method for calculating CNR in the images from digital mammography systems, using a single ROI in the aluminium square and a single ROI in the background. In this work, the relative noise was calculated using this method for a range of ROI sizes using an image from the Agfa CR system. The results showed that ROI size affects the relative noise measurement and heel effect correction is needed or a small ROI size should be used. If a small area of ROI is used, multiple small ROIs will improve the statistics by obtaining a larger data sample. The use of a single background ROI with a non-heel effect corrected CR image is likely to lead to inaccuracies in the contrast measurement and therefore the CNR measurement. This additional problem was not quantified in this work as the CNR measurements presented all used four background ROIs surrounding the aluminium square.

4.4.3 Impact of ROI Size and Heel Effect on CNR Measurement

CNR was calculated for images from the CR systems using a range of ROI sizes and doses. The results showed significant dependence of the CNR measurement on ROI size. After applying the heel effect correction to the images, the CNR was recalculated using 2 x 2 cm ROI as this is the ROI size specified in the European Guidelines for the CNR measurement. It was found that using 2 x 2 cm background ROI in the images with and without heel effect correction, resulted in different CNR values. This was observed with all CR systems used here but it was more significant with the Konica CR system. It is likely that the magnitude of the error depends on the magnitude of the heel effect on the X-ray set used rather than on the specific type of CR system involved, which themselves have a relative uniform response. The difference in the CNR values measured with and without heel effect correction was calculated as in Eq.4.7 and the results are summarised in Table 4.3 for all CR systems.
CHAPTER 4. RELATIVE NOISE AND CNR MEASUREMENTS

\[
\text{Error} \, (\%) = \frac{\text{CNR}_{\text{corr}} - \text{CNR}}{\text{CNR}_{\text{corr}}} \times 100
\]

(4.7)

where \( \text{CNR}_{\text{corr}} \) is the CNR calculated in the heel effect corrected image using a 2 × 2 cm ROI, and \( \text{CNR} \) is the CNR calculated in the image before correction.

It can be seen from Table 4.3 that the difference in the CNR values calculated with and without heel effect correction decreases as the ROI size decreases and becomes very small (<1%) at the ROI size of 0.25 × 0.25 cm. This means that the use of multiple 0.25 × 0.25 cm ROIs gave a result that was essentially the same as if a heel effect correction had been applied. The heel effect correction method requires an extra image for each CNR measurement, whereas the use of multiple small ROIs has the advantage that only a single image is required.

Table 4.3 CNR values calculated, at the AEC selected dose, for different ROI sizes for the Agfa, Kodak and Konica CR systems. The CNR values measured from the 2 × 2 cm ROI in the heel effect corrected image are also shown.

<table>
<thead>
<tr>
<th>ROI size (cm)</th>
<th>Heel effect</th>
<th>CNR (Agfa)</th>
<th>CNR (Kodak)</th>
<th>CNR (Konica)</th>
<th>Error (Agfa)</th>
<th>Error (Kodak)</th>
<th>Error (Konica)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2x2</td>
<td>Yes</td>
<td>13.64</td>
<td>10.13</td>
<td>8.24</td>
<td>0.0%</td>
<td>0.0%</td>
<td>0.0%</td>
</tr>
<tr>
<td>2x2</td>
<td>No</td>
<td>12.29</td>
<td>9.32</td>
<td>6.70</td>
<td>10%</td>
<td>8.0%</td>
<td>18.3%</td>
</tr>
<tr>
<td>1x1</td>
<td>No</td>
<td>13.09</td>
<td>9.63</td>
<td>7.86</td>
<td>4.0%</td>
<td>4.9%</td>
<td>4.6%</td>
</tr>
<tr>
<td>0.5x0.5</td>
<td>No</td>
<td>13.30</td>
<td>9.88</td>
<td>8.14</td>
<td>2.5%</td>
<td>2.5%</td>
<td>1.21%</td>
</tr>
<tr>
<td>0.25x0.25</td>
<td>No</td>
<td>13.61</td>
<td>10.04</td>
<td>8.23</td>
<td>0.2%</td>
<td>0.88%</td>
<td>0.12%</td>
</tr>
</tbody>
</table>

4.5 Conclusions

The impact of the ROI size on the relative noise and CNR measurements was investigated for CR and DR systems. The measured relative noise for the CR images strongly depended on the ROI size due to the heel effect. After applying the heel effect correction there was very little dependence on ROI size. The relative noise in the images from the DR system also showed very little dependence on ROI size – presumably due to the flat-field correction applied by the manufacturer. However, the results suggest that the use of multiple very small ROIs produces a noise measurement that is close to that found after applying a heel effect correction. The effect of ROI size and heel effect correction was found to have a corresponding impact on the measurement of CNR. In this case the heel effect distorted the CNR measurement when larger ROI was used. However, the use of multiple very small ROIs
CHAPTER 4. RELATIVE NOISE AND CNR MEASUREMENTS

ROIs led to a result that was essentially the same as if a heel effect correction had been applied. Thus the appropriate ROI size which can be used for CNR measurement without the need for the heel effect correction was found to be 0.25 cm × 0.25 cm. With this size and at the clinical exposure conditions the heel effect had an insignificant impact on the measurement of relative noise and CNR. The use of multiple small ROIs to determine the contrast signal and background noise has the advantage that only a single image is required. The application of a heel effect correction for measurements with CR systems requires two images and some complex image processing. The current suggestion in the European guidelines to use a 2 × 2cm ROI is not suitable for CR systems and leads to an error of 8% to 18% in CNR determination due to the heel effect. Correcting for such errors in CNR measurement will be particularly important where CNR is used to optimise a CR system as described previously [4,8].

References


CHAPTER 4. RELATIVE NOISE AND CNR MEASUREMENTS

Chapter 5

Evaluation of Automatic Reading of CDMAM Test Object

5.1 Introduction

The European guidelines for the quality control of mammography include minimum standards for the image quality of digital mammography systems [1]. These standards are based on contrast-detail measurements and the work undertaken to develop these minimum standards has been described previously [2]. The method involves the determination of threshold contrast visibility using detail sizes from 2.0 mm down to 0.1 mm using the contrast detail mammography (CDMAM) phantom (version 3.4, Artinis, St. Walburg 4, 6671 AS Zetten, The Netherlands). The minimum standards were set to ensure that digital systems are as good or better than film screen systems.

In practice contrast detail measurements should rely on a large number of observer readings. This procedure suffers from two main disadvantages. One is the presence of significant inter-observer error which undermines the reliability and confidence in the measurements. The other disadvantage is that using human observers is time consuming. It would therefore be desirable to have an automatic method of obtaining threshold contrast data when testing digital mammography systems. A possible solution to these problems is software, which allows automatic reading of CDMAM images. Karssemeijer and Thijssen described the use of such software to determine threshold contrasts [3]. The basic software tool described by Karssemeijer and Thijssen for automatically identifying discs on digital images of the CDMAM is

* Published in Proceedings of SPIE Medical Imaging (2008), vol. 6913, 6913-1C-11, with minor modifications
called CDCOM and is available for downloading at the EUREF website [4]. Also provided at this website is a manual explaining how to use the software [4].

The CDCOM programme attempts to correctly locate the position of the gold discs on a single DICOM image. However the programme does not combine the data from more than one image or determine the threshold contrasts. A method for doing so was described by Karssemeijer and Thijssen [3] and Veldkamp et al [5]. Thus CDCOM appears to be a potentially useful tool for determining threshold contrasts. Fletcher-Heath and Van Metter have described an alternative approach to using the CDCOM data to determine threshold contrasts from CDMAM images [6]. The manufacturers of the CDMAM test object also offer for sale a commercial software package which uses CDCOM to determine threshold contrasts from a set of 8 CDMAM images. However, although this software uses CDCOM the method of determining the threshold contrasts is different from those described previously [3,5,6]. One limitation of the first publications using CDCOM was that the relationship between automatic and human observer scoring was not fully explored across the wide variety of systems and circumstances encountered in practice. All the publications report that automatic analysis yields lower threshold contrasts than human observers. Young et al had reported on the relationship between human and automatic readings [7] and described a method of predicting the threshold contrast observed by a typical human observer [8]. This, however, was based on a small number of readers and a small population of images provided by the National Coordinating Centre for the Physics of Mammography (NCCPM) in Guildford.

The aim of this chapter is to further investigate the use of CDCOM (Version 1.5) to read images of the CDMAM test object using a large number of readers and a large population of images provided by three different centres\(^\ast\). In general, the following aspects were studied in details:

- the ratio of automated readings and human readings at three different centres.
- the reproducibility of automatic reading using different numbers of images and for different types of system.

\(^\ast\)The three centres are Guildford, UK; Nijmegen, Netherlands; and Leuven, Belgium.
• the variation in automatic readings between test objects with different serial numbers

However, the overall objective is to develop a reliable automatic method of assessing digital mammography systems against European Guidelines.

5.2 Materials and Methods

5.2.1 Systems Tested and Image Acquisition

Sets of 113 CDMAM images (8 images each) were acquired using a wide range of digital mammography systems. The acquired images were then saved as unprocessed (raw) DICOM files for later analysis. The different systems used are shown in Table 5.1. The images were obtained at three centres (Guildford, UK; Nijmegen, Netherlands; and Leuven, Belgium). Some of the systems were operated at a variety of doses and beam quality and some had additional sets of CDMAM images so that the reproducibility of the automatic analysis could be measured. The tube voltage target/filter combinations used in producing the images of the CDMAM test object are also shown in Table 5.1.

The CDMAM phantom (version 3.4) was radiographed on each digital mammography system used. The CDMAM consists of a matrix of gold discs of thicknesses from 2μm to 0.03μm and diameters from 2mm to 0.06 mm on a 0.5 mm aluminium base encased in PMMA. The assembly (PMMA and aluminium) has a Plexiglas-equivalent thickness of 10mm. The phantom was positioned with a 20 mm thickness of PMMA blocks above and below. This combination has a total attenuation approximately equivalent to 50mm of PMMA. This has been shown to be equivalent to breasts of typical composition with a compressed thickness of 60 mm. Expanded polystyrene spacers were added at the edges of the phantom to create a total thickness of 60 mm and a standard 100N compression applied. This arrangement was imaged using the factors automatically selected by the X-ray set and these were recorded. The phantom was repositioned very slightly between exposures (i.e. about a 1mm shift) and the exposures repeated until a set of 8 CDMAM images were obtained. If the error in the threshold determination was to be estimated a further similar 8 CDMAM images were obtained.
CHAPTER 5. CDMAM AUTOMATIC READING

To assess the effect of dose on threshold contrasts further sets of CDMAM images were obtained on some systems (at the Guildford Centre) by manually selecting mAs that were approximate multiples of 2 higher or lower (i.e. double and half dose) than selected using the AEC control. The tube voltage and target/filter combinations were kept the same.

Table 5.1. Systems used in this work

| CR1 | Agfa CR75.0 | 50 | Siemens Mammonat 3000 | 28/MoMo |
| CR2 | Agfa CR Embrace | 50 | GE Senographe DMR+ | 30/MoMo |
| CR3 | Agfa CR85.0 | 50 | Hologic LoRad | 28/MoMo |
| CR4 | Fuji Profect CS | 50 | GE Senographe DMR+ | 28/MoMo |
| CR5 | Fuji Profect CS | 50 | Siemens Mammonat 3000 | 27/MoRh |
| CR6 | Fuji Profect CS | 50 | Planmed | 30/MoMo |
| CR7 | Fuji Profect CS | 50 | Hologic LoRad | 28/MoMo |
| CR8 | Fuji Profect CS | 50 | Ge Performa | 30/MoMo |
| CR9 | Fuji 5000MA | 50 | Siemens Mammonat 3000 | 28/MoMo |
| CR10 | Kodak Directview CR 850 | 100 | Siemens Mammonat 3000 | 27/MoRh |
| CR11 | Kodak Directview CR 850 | 50 | GE Senographe DMR+ | 27/MoRh |
| CR12 | Kodak Directview CR 950 | 50 | GE Senographe DMR+ | 27/MoRh |
| CR13 | Kodak Directview CR 975 | 50 | GE Senographe DMR+ | 27/MoRh |
| CR14 | Konica CR Regius 190 | 43.75 | GE Senographe DMR+ | 27/MoRh |
| CR15 | Philips PCR ElevaCosimaX | 50 | Siemens Mammonat 3000 | 28/RhRh |
| DR1 | GE Senographe 2000D | 100 | n/a | 28/RhRh |
| DR2 | GE Senographe DS | 100 | n/a | 29/RhRh |
| DR3 | GE Essential | 100 | n/a | 29/RhRh |
| DR4 | Hologic Selenia | 70 | n/a | 29/MoMo |
| DR5 | IMS Giotto Image MD | 85 | n/a | 31/MoRh |
| DR6 | Agfa DM1000 | 70 | n/a | 29/MoMo |
| DR7 | Fischer Senovation | 48 | n/a | 29/W Al |
| DR8 | Siemens Novation | 70 | n/a | 28/W Rh |
| DR9 | Planmed Nuance | 85 | n/a | 29/MoMo |
| DR10 | Sectra Microdose | 50 | n/a | 32/W Al |
| DR11 | Sectra D40 | 50 | n/a | 35/W Al |

5.2.2 Dose Measurements

For each exposure the factors used when imaging the CDMAM phantom with the additional PMMA were recorded. The X-ray setting output, half-value layer (in mm of aluminium) and the distance from the focus to Table top were measured allowing the entrance surface air kerma at the top of a 50mm thickness of PMMA to be calculated. The method described by Dance et al [9] was used to calculate the mean glandular dose (MGD) to typical breasts with a 60 mm compressed breast thickness and an attenuation equivalent to a 50 mm thickness of PMMA. For each exposure the MGD of a breast equivalent to the test phantom was calculated as,

* Some systems were used with additional tube voltage target/filter combinations
where $K$ is the entrance surface air kerma, $g$ is the conversion factor which relates $K$ to MGD for the typical compressed breast and depends on radiation quality, $c$ is the correction factor for the difference in breast composition from 50% glandularity and $s$ is the factor that corrects for using other than original target/filter combination (Mo/Mo) [10]. The corresponding conversion factors used here are tabulated in [9] and [10]. The average of the MGD values for each set of similar CDMAM images was then calculated.

5.2.3 Human Readings

All the raw CDMAM images were read by the human observers at the centre where the images were obtained. The scoring of the CDMAM phantom images is performed by indicating the corner of each square in which a disc appears to be present. For the Guildford centre, the observers work in the same imaging laboratory of the Medical Physics Department, Royal Surrey County Hospital (RSCH), Guildford. The laboratory had a major role in establishing the limiting values in the European protocol. The observers had received training and practice in scoring CDMAM images from experienced observers. Observers were instructed to guess the disc location for at least one square beyond what they can see. This training is important as new observers take a little practice until they can see the discs to their optimal performance. During the training period an observer who fails to see discs that other observers can see is asked to try again until they reach a similar level of performance. Images scored for training purposes are not included in this study. Occasionally an observer cannot be trained to see the discs to a level consistent with the other observers and is not then further engaged in the study.

The digital CDMAM images were displayed on a viewing station using a diagnostic quality 3 Mega Pixel DICOM calibrated display monitor. The scoring was performed in a very dark ambient. The contrast and brightness of each image was adjusted to optimally display the details in the test object, before scoring. The observer could use as much electronic zoom as needed and background illumination was kept to a minimum. The scoring of the CDMAM images was performed using the method described in chapter 2. The manual for the CDMAM phantom explains how to apply a
nearest neighbour correction (NNC) rules to the scores for each reading of a CDMAM image [4]. These NNC rules were applied to each of the images read by a human observer in this study. After applying these rules the smallest (threshold) gold thickness for a correctly indicated disc was noted for each diameter. The standard procedure used here was to have three observers each score 4 images from a set of CDMAM images and to use these to find the average threshold gold thickness for each diameter (in the European protocol it is suggested that 3 observers score only 2 different images each [1]).

For the other two centres (Nijmegen and Leuven), the raw CDMAM images were read by human observers at their centre. All the human observations were processed at one centre (the Guildford Centre) to determine threshold gold thickness, to ensure the method of analysis was consistent. This was done by transferring the electronic (Excel spreadsheet) completed score forms CDMAM-phantom to the author in the Guildford centre. The score forms were then converted into a standard method to enable the NNC rules. This was done using appropriate functions (macros) in the Excel spreadsheets. Because each centre has its own method to score the CDMAM images, different macros were used. The NNC rules were applied automatically using a special macro in the Excel spreadsheets. Figs.5.1 and 5.2 show the original score forms of the Nijmegen and Leuven centres, respectively. Fig.5.3 shows the converted score form before and after the NNC rules for the Nijmegen centre.

<table>
<thead>
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<th>Diameter [mm]</th>
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<th>0.04</th>
<th>0.05</th>
<th>0.06</th>
<th>0.08</th>
<th>0.10</th>
<th>0.13</th>
<th>0.16</th>
<th>0.20</th>
<th>0.25</th>
<th>0.36</th>
<th>0.50</th>
<th>0.71</th>
<th>1.00</th>
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</tr>
</thead>
<tbody>
<tr>
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<td>G</td>
<td>G</td>
<td>G</td>
<td>G</td>
<td>G</td>
<td>G</td>
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<td>F</td>
<td>F</td>
</tr>
</tbody>
</table>

Fig.5.1. The original score form CDMAM-phantom (in Excel sheet) obtained from Nijmegen centre. G: disk is correctly indicated, F: disk is incorrectly indicated and N: disk is absolutely not indicated.
CHAPTER 5. CDMAM AUTOMATIC READING

Fig. 5.2. The original score form CDMAM-phantom (in Excel sheet) obtained from Leuven centre. The numbers scored represent the corner in which a disk appears to be present.

Fig. 5.3. The score form CDMAM-phantom (in Excel sheet) of the Nijmegen centre (a) before and (b) after the NNC rules. The smoothing is indicated by the thick red line. The threshold gold thickness and the threshold diameter are also shown.

80
After applying the NNC rules the smallest (threshold) gold thickness for a correctly indicated disc was noted for each diameter, as shown in Fig. 5.3b. The procedure used in these two centres was to follow the European protocol which suggests that 3 observers score only 2 different images each from a set of CDMAM images and to use these to find the average threshold gold thickness for each diameter.

The threshold contrast $C_T$ was converted from a gold thickness $T$ using a nominal radiation contrast $C$ (%) and a beam hardening coefficient $h$ for an X-ray spectrum of 28 kVp Mo/Mo and 45 mm PMMA, as per Eq. 5.2. This equation was found by fitting a polynomial fit to a set of known results from that X-ray spectrum using published data [11], as described in the European protocol. From the fitting it was found that $C$ and $h$ equal 15.73% and 1.18, respectively.

$$C_T = C \cdot T + h \cdot T^2$$  \hspace{1cm} (5.2)

The average threshold contrast for each detail diameter for each system was fitted with a curve of the form shown in Eq. 5.3.

$$TcF = a + b \cdot x^{-1} + c \cdot x^{-2} + d \cdot x^{-3}$$  \hspace{1cm} (5.3)

where $TcF$ is the fitted threshold contrast (%) calculated at 28 kVp and Mo/Mo target/filter combination, $x$ is the detail diameter (mm), and $a$, $b$, $c$ and $d$ are the coefficients adjusted to achieve a least square fit. This was done using the Solver tool within an Excel spreadsheet.

### 5.2.4 Automatic Readings

All the sets of CDMAM images were read automatically in the Guildford centre using the CDCOM and the method described by Karssemeijer and Thijssen [3] and Veldkamp et al [5]. In this method, all the 16 detection matrices were combined to produce a detection matrix showing the fraction of discs correctly detected in cells corresponding to the cells in the phantom, as shown in Fig. 5.4. The method of determining the threshold gold thickness is to fit a psychometric curve for each detail diameter as described in Eq. 5.4.
where \( p(t) \) is the probability of detecting a gold disc of certain size with a thickness \( t \), \( C \) is the logarithm of radiation contrast of the gold disk \( C = \log(l - e^{-\mu t}) \), \( f \) is a free parameter to be fitted and \( C_T \) is the threshold contrast corresponding to \( p = 0.625 \) which is the mid-point between completely correct scoring (\( p = 1.00 \)) and random guessing (\( p = 0.25 \)).

For the linear attenuation of gold, \( \mu = 0.190 \, \mu m^{-1} \) was used. A fixed value of \( f \) was used for all diameters, as described previously [5]. The curve fitting was a least mean squares procedure accomplished using the Solver tool within an Excel spreadsheet. The threshold gold thickness is taken as the point on the fitted curve where \( p = 0.625 \). This curve fitting procedure is repeated to provide threshold gold thicknesses for each detail diameter as shown in Fig.5.5. These threshold gold thicknesses were converted to threshold contrast as described earlier. The data for the 0.06mm and 0.08mm detail diameters are usually not used because the psychometric curve fit could not be fitted accurately because all values of \( p \) may be below 0.625.

Consequently, CDCOM could not effectively locate the discs for detail diameters of less than 0.1mm. In effect, the measurements have gone off the scale of the phantom range. A similar problem occurs for large details where all \( p \) values may be much
greater than 0.625. In other words, CDMAM easily locates almost all the discs for large diameters. This is why the threshold contrasts are not used for the 1.6mm and 2.0mm detail sizes. However, the psychometric curve generally provided a good fit for details with diameters from 0.1mm to 1.0mm and this is the range used for the results shown here.

![Psychometric curve fit](image)

Fig. 5.5. An example of psychometric curve fit (lines) to the detection matrix data of (dots) for detail sizes from 0.13mm to 0.2 mm. In this case the detection matrix was smoothed before curve fitting.

In order to optimize the procedure, four different methods of determining the threshold gold thickness were used for every set of measurements, as described in Table 5.2. Where no specific method is specified in the results, method D has been used. Where a contrast-detail curve was fitted Eq. 5.3 was used.

**Table 5.2. Variations in the method of determining the threshold gold thickness**

<table>
<thead>
<tr>
<th>Method</th>
<th>Detection matrix smoothed before fitting psychometric curve</th>
<th>Contrast-detail curve fitted and fitted values used instead of raw values</th>
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</thead>
<tbody>
<tr>
<td>A</td>
<td>No</td>
<td>No</td>
</tr>
<tr>
<td>B</td>
<td>No</td>
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<tr>
<td>C</td>
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<td>No</td>
</tr>
<tr>
<td>D</td>
<td>Yes</td>
<td>Yes</td>
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</tbody>
</table>
CHAPTER 5. CDMAM AUTOMATIC READING

The method used to smooth the detection matrix was described previously [2] and summarised here. In this method the 16 matrices were added together and a simple 3×3 smoothing algorithm was applied to the detection matrix (such as the matrix in Fig.5.4). Each cell was replaced by the weighted average of itself and the adjacent cells using the relative weightings shown below.

\[
\begin{pmatrix}
1 & 2 & 1 \\
2 & 4 & 2 \\
1 & 2 & 1 \\
\end{pmatrix}
\]

Smoothing algorithm

This smoothing has the effect of reducing the random fluctuations in the cell values making interpolation easier. Corrections to the weighting were made at the edges of the matrix. The smoothing algorithm was implemented using Excel spreadsheet functions. Fig.5.6 shows the smoothed matrix of the detection matrix demonstrated in Fig.5.4.

<table>
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<th>0.08</th>
<th>0.10</th>
<th>0.13</th>
<th>0.16</th>
<th>0.20</th>
<th>0.25</th>
<th>0.30</th>
<th>0.36</th>
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<th>0.71</th>
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Fig.5.6. An example of proportion correctly detected matrix after smoothing. The detected matrix demonstrated in Fig.5.4 was used.

5.2.5 Automatic to Human Ratio

The correlation between the results of the automated process and the human readings was determined and compared for the three centres. In each case a non-linear correlation was measured using a power function of the form shown in Eq. 5.5.

\[ T_{C_{human}} = a[T_{C_{auto}}]^n \] (5.5)
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where $T_{C_{\text{human}}}$ is the average threshold contrast for a team of human observers, $T_{C_{\text{auto}}}$ is the average threshold contrast determined using the automated programme, $a$ and $n$ are coefficients to be fitted.

The average ratio of human to automatic threshold contrasts was determined at each detail size to show the difference in the performance of the human observers between the centres. To study whether the ratio of human to automatic threshold contrast varied from one system type to another, Eq. 5.6 was used to predict the human readings at each centre.

$$T_{C_{\text{predicted}}} = a[T_{C_{\text{auto}}}]^n$$ (5.6)

where specific values of the coefficients $a$ and $n$ were used for each of the three centres. The reason for adopting this approach was to minimise confounding by differences between the readers at the three centres. The average ratio of predicted threshold contrast over actual average threshold contrast measured by the human observers was then calculated for each image set. From these ratios the average ratios for different types of systems were calculated.

5.2.6 Reproducibility of Automatic Readings

Three large sets of images (64 images) of one CDMAM test object with similar exposure settings were obtained on three different systems (GE Senographe 2000D, Planmed Nuance, Agfa CR 85-X with MM3.0 plate). A set of 8 images was randomly selected from each pool of images and automatically analyzed. This was done using a code written in the C++ programme (Appendix A). The random selection was repeated 150 times producing 150 fraction correctly detected matrices. These fractions were then processed automatically, using a code written in the Visual Basic programme in Excel spreadsheet (appendix A), to determine the threshold contrast at each detail diameter. The reproducibility (coefficient of variation) of each threshold contrast calculation was then obtained as,

$$\text{Coefficient of variation} = \frac{SD(\text{threshold contrast})}{\text{Mean(threshold contrast)}}$$ (5.7)
The coefficient of variation was calculated for the three digital mammography systems and the results averaged. The effect of increasing the number of images used in automatic analysis was also assessed by selecting larger sets of images (i.e. using 16 and 32 instead of 8 images).

5.2.7 Comparison of test objects

Four CDMAM test objects were imaged 32 times using similar exposure factors on the Planmed Nuance system. The images of each test object were automatically analyzed to determine the effect of individual test objects on the measurements of threshold gold thickness.

5.3 Results

5.3.1 Human Readings

The numbers of sets of CDMAM images read by the human observers at each centre are shown in Table 5.3. The normal number of human readings per image set varied between the centres as shown in Table 5.3. The measured human threshold contrasts were obtained for the details of 0.1 – 1.0 mm for all systems used. Fig.5.7 shows the smoothed human threshold contrasts and the fitting curves plotted as a function of details, for the GE Essential mammography.

<table>
<thead>
<tr>
<th>Centre</th>
<th>Number of sets of CDMAM images</th>
<th>Typical number of human observations per set of images</th>
</tr>
</thead>
<tbody>
<tr>
<td>Guildford</td>
<td>69</td>
<td>12</td>
</tr>
<tr>
<td>Nijmegen</td>
<td>29</td>
<td>8</td>
</tr>
<tr>
<td>Leuven</td>
<td>21</td>
<td>4</td>
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</table>

5.3.2 Automatic Readings

The measured automated threshold contrasts were calculated using method A, for the details of 0.1 – 1.0 mm for all systems used. Fig.5.8 shows the threshold contrasts plotted as a function of details for the Hologic selenia mammography. Also shown the threshold contrasts after applying curve fitting using method B. Fig.5.9 shows the smoothed threshold contrasts (method C) for the system plotted as a function of details of 0.1 – 1.0 mm. The smoothed threshold contrasts after applying the curve fitting (method D) are also shown.
Fig. 5.7. Smoothed measured human threshold contrasts of the GE Essential system (at MGD of 2.04 ± 0.43 mGy). Error bars indicate 95% confidence limits. The minimum acceptable and achievable standards in the European protocol are also shown.

Fig. 5.8. Automated threshold contrast (method A) of the Hologic sezenia digital mammography system (at the AEC selected dose). The fitted threshold contrasts (method B) are shown. The minimum acceptable and achievable standards in the European protocol are also shown.
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5.3.3 Auto to Human Ratio

The correlation between the results of the automated method and the human readings was determined and compared for each centre using Eq.5.5. Fig.5.10 shows scatter plots of the threshold contrasts determined by human observers and the automated programme for the three centres. The threshold contrasts plotted for the human readings represent the average for all the readings at that centre. The correlation between human and automatic determinations are shown for each of the centres in Figs 5.10a, b and c. Fig.5.11 shows the average ratio of human to automatic threshold contrasts for the three centres. The curves shown for each centre in Fig. 5.11 show the average ratios when the correlations found for each centre in Fig.5.10 are used to predict the human threshold contrast from the automated readings using Eq.5.6.

5.3.4 Reproducibility of Automatic Readings

The coefficients of variation for the four methods of determining threshold gold thickness are shown in Fig.5.12 for one of the systems. The coefficients of variation

Fig.5.9. Automated threshold contrast after smoothing process (method C) of the Hologic selenia digital mammography system (at the AEC selected dose). The fitted threshold contrasts (method D) are shown. The minimum acceptable and achievable standards in the European protocol are also shown.
for automated measurement of threshold gold thickness on three different systems are shown in Fig. 5.13. The effect of increasing the number of images used in the automated measurement is shown in Fig. 5.14. The measurements in the two figures were made using method D.

![Graphs of human and automatic threshold contrasts for different centers](image)

Fig. 5.10. Scatter plots of human and automatic threshold contrasts for (a) Guildford, (b) Nijmegen and (c) Leuven centres. The scatter plots for all three centres superimposed in plot (d). Errors on coefficients are ± 1 standard error.
Fig. 5.11. Ratio of human to automatic threshold contrasts for Guildford, Nijmegen and Leuven centres. The nearest neighbour correction rules were applied to the human readings at the Guildford centre. Error bars indicate ±2 standard errors of the mean. Note that this figure includes data from all the systems.

Fig. 5.12. Reproducibility of the different methods of determining threshold gold thickness for a set of 8 CDMAM images using an Agfa CR system. Error bars indicate ±2 standard errors of the mean.
Fig. 1.13. The coefficients of variation for automated measurement of threshold gold thickness on three different systems. Error bars indicate ±2 standard errors of the mean. Note: method D was adopted in this measurement.

Fig. 5.14. The coefficients of variation for automated measurement of threshold gold thickness using different numbers of images. Error bars indicate ±2 standard errors of the mean. Note: method D was adopted in this measurement and Planmed system was used.
5.3.5 Comparison of test objects

The automatically determined threshold gold thicknesses using 4 different test objects are compared in Fig.5.15.

![Graph showing comparison of test objects](image)

Fig.5.15. Comparison of the contrast detail curves for the same system using 4 test objects with different serial numbers. 32 images were used per test object using Planmed system. Error bars indicate ± 2 standard errors of the mean.

5.4 Discussion

5.4.1 Human and Automatic Reading

The human readings at each centre showed a good correlation with the automated readings. However, these correlations were different. This could most easily be seen in Fig.5.11, which showed that the average reader at Leuven measured lower threshold contrasts than the average reader in Guildford. The readers in Nijmegen were between these two. There may be a number of reasons for these differences but the most obvious is inter-observer differences known to exist between observers conducting such a task. For practical purposes it would be important to agree on what represents the typical relationship between human and automatic readings. However,
by averaging over the readers of all three centres it was possible to obtain a better approximation of the average reader.

The ratio between predicted and human threshold contrasts was analysed by system type in Table 5.4. If this ratio is significantly different from 1.0 for any system type it suggests that the predicted values were either an under or over estimate of true human readings. For almost all diameters and detail sizes considered, the ratio was not significantly different from 1.0. However, for the 0.1mm detail size the ratio was 0.93 ± 0.05 (2 sem) for the systems using the Hologic selenium detector. This suggests that for these systems the predicted values may be about 7% lower than typical human readings. Similarly for the five Sectra systems the average ratio for the 0.1mm details was 1.16 ± 0.08. This implies that for these systems the predicted values may be about 16% higher than typical human values. This merits further investigation to determine whether this is a consistent pattern and whether there are related factors.

5.4.2 Reproducibility of Automatic Readings

Four methods of determining the threshold gold thickness were used in this work. Fig. 5.12 showed that method D was the most reproducible when using 8 images with the Agfa CR system. In fact, this was a general finding for the three types of systems and for different numbers of images. The application of smoothing and curve fitting to the data could have changed the mean values as well as improving the reproducibility. However, it seems from Table 5.5 that this has not happened to any significant degree. Thus method D produced the same average threshold gold thickness as method A. Therefore, method D was selected as the standard procedure and used in the rest of the results presented.

<table>
<thead>
<tr>
<th>Mammography system</th>
<th>No of measurements</th>
<th>Ratio of predicted to human threshold contrasts (± 2 SEM)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>0.1 mm</td>
</tr>
<tr>
<td>All systems</td>
<td>113</td>
<td>0.98 ± 0.03</td>
</tr>
<tr>
<td>CR</td>
<td>60</td>
<td>1.00 ± 0.03</td>
</tr>
<tr>
<td>DR (all types)</td>
<td>53</td>
<td>0.96 ± 0.04</td>
</tr>
<tr>
<td>DR (GE)</td>
<td>11</td>
<td>0.93 ± 0.08</td>
</tr>
<tr>
<td>DR (Selenia)</td>
<td>24</td>
<td>0.93 ± 0.05</td>
</tr>
<tr>
<td>DR (Sectra)</td>
<td>5</td>
<td>1.16 ± 0.08</td>
</tr>
</tbody>
</table>
When 8 images were used the coefficient of variation was about 4% for three different types digital mammography systems for details with diameters between 0.2 and 1.0 mm. Below 0.2 mm the variation was higher and was about 10% at 0.1 mm. The higher coefficient of variation for the smallest detectable discs (typically 0.1 mm) could be explained by a number of factors. One explanation is that the 0.1 mm disc is often the smallest point in the contrast detail curve. Consequently there are no threshold contrast values for smaller discs, so curve fitting does not improve the accuracy as it does for the larger discs. The 0.1 mm disc is relatively close to the detector element (pixel) sizes used in digital mammography i.e. 50 to 100 pm. This means that the contrast signal may be wholly detected by one pixel or distributed across several. This random effect may cause the detectability of such small discs to be more variable i.e. a partial filling effect. Finally, the measured threshold contrast for the 0.1 mm discs is usually close to the edge of the phantom. This means that the fitting of the psychometric curve for this diameter may be less accurate as there are fewer useful data points to be fitted. A similar problem occurs for details greater than 1 mm in diameter. In this case almost all the details are detected and curve fitting becomes inaccurate. Thus, it seems that some of these sources of error are related to the limitations of the phantom design and could be reduced with a different design.

However, using larger numbers of images improved the reproducibility of the process for all detail sizes. While it seems inappropriate to use such large numbers of images for routine quality control, it is practical for type testing and design evaluation.
5.4.3 Comparison of test objects

One of the four test objects, CDMAM 1022, produced consistently lower threshold gold thicknesses than the other three. The magnitude of this difference was about 10%. The other three phantoms produced threshold gold thicknesses that were in agreement within the measurement error of about 2% (1 standard deviation) between 0.1mm and 0.5mm detail sizes. CDMAM 1210 diverged from the other two test objects, above 0.5mm detail size, by up to 10%.

5.5 Conclusions

The ratio of automatic to human determinations of threshold contrast has been quantified at three different centres. These data provide a means of predicting average human performance using the automated reading software. The coefficient of variation in automatically determined threshold gold thickness was about 4% for detail sizes from 0.2 to 1.0mm when 8 images were analysed. The coefficient of variation was about 10% at a detail size of 0.1mm. Using larger numbers of images and a change in phantom design could greatly improve reproducibility. Greater consistency of phantom construction would also be desirable as one of the four phantoms tested was significantly different from the other three. Despite some limitations automated reading of CDMAM images can provide a reproducible means of assessing digital mammography systems against European Guidelines.

References


Chapter 6

Evaluation of MTF Measurements

6.1 Introduction

The image resolution properties of a system are most often described by its modulation transfer function (MTF) [1]. MTF describes the ability of an imaging system to transfer the signal from an object to the image over a range of spatial frequencies. Different methods were developed to determine the MTF of a mammographic system including edge [1] and slit [2] methods. Also the international electrotechnical commission (IEC) has recently published a protocol for measuring MTF, NPS and DQE of the detectors used in mammography [3]. In the literature, the impact of system factors on the MTF has been evaluated for general radiography [4]. The physical characteristics of a prototype imager have also been investigated under various clinically relevant mammographic spectral conditions [5]. Moreover, the performance, in particular the DQE, of a clinical digital mammography system has been previously characterized at a range of spectral conditions [6].

In this chapter, the MTF measurements of widely used clinical DR and CR digital mammography systems are evaluated under various spectral and clinical conditions. These include tube voltage and current, target/filter combination, type, position and thickness of added filtration, presence of compression paddle and anti scatter grid, and thickness, orientation and position of the edge device.

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6.2 Materials and Methods

6.2.1 Systems Tested and Image Acquisition

The study was carried out using two CR systems and two DR systems: Fuji Capsula CR, Konica Minolta Regius 190 CR, GE Essential, and Giotto IMS, respectively. The CR systems use the photostimulable storage phosphor technique. The Konica system was used with three different types of mammography CR plate. The RP-6M plate is a conventional powder plate that was designed previously for use with the system. The RP-7M is an improved version of the RP-6M. The CP-1M is a new type of plate with a phosphor that employs needle crystal technology designed to improve the sharpness of the images by reducing light spread in the phosphor layer.

The GE system is an indirect-conversion flat-panel consists of amorphous silicon and caesium iodide activated with thallium [CsI(Tl)]. The Giotto system is a direct conversion flat-panel made of amorphous selenium photoconductor which converts the X-ray directly into electric charge. Table 6.1 summarises the detector and image specifications for each unit used in the study.

Table 6.1. Physical characteristics of the Systems used in the study

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>Model</th>
<th>Pixel size (μm)</th>
<th>Mammography X-ray set</th>
<th>Image plate</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE Medical Systems</td>
<td>Essential</td>
<td>94</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Giotto IMS</td>
<td>Image MD</td>
<td>80</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Fuji Photo Film</td>
<td>FCR Capsula</td>
<td>50</td>
<td>GE DMR</td>
<td>HR-BD</td>
</tr>
<tr>
<td>Konica Minolta</td>
<td>Regius 190</td>
<td>43.75</td>
<td>GE DMR</td>
<td>RP-6M</td>
</tr>
</tbody>
</table>

Each measurement was performed at a hospital or centre where the unit had been installed. All images acquired during the measurement were saved as unprocessed files in DICOM format. These are then transferred to a personal computer for later analysis.

6.2.2 Detector response and linearity

To assess a medical imaging system in terms of MTF, the system has to be linear and shift invariant [7]. It is well known [8] that digital detectors used in the DR systems have a linear response to the input radiant exposure. On the other hand, CR systems have a non-linear response and hence need to be linearised before measuring their
MTF [8]. Contrary to conventional detectors (i.e. film), digital detectors are shift-variant systems which mean that their MTF depends on the position of the input radiation source, i.e. phase dependent. The difficulties in the MTF measurements for the shift-variant detectors and the solutions to these difficulties were discussed at length in chapter 2.

The detector response of each system was measured broadly as described in the UK protocol [9]. Fig. 6.1 shows the experimental setup for this measurement. Uniform images were acquired for the same exposure conditions for a range of mAs settings. The X-ray beam was filtered using 45 mm Plexiglas (PMMA) mounted on the X-ray tube. PMMA of this thickness was used to simulate the attenuation due to a real compressed breast [10] of typical composition (29% glandularity) with a compressed thickness of 53 mm [11]. The detector entrance air kerma was measured using an MDH 1015C X-ray monitor and a 10 x 5-6M dedicated mammography ionisation chamber (RadCal Corp., Monrovia, CA). The chamber was positioned at the centre of the surface of the breast support Table. The readings were corrected to the surface of the imaging detector (the cassette in the CR systems) using the inverse square law. The measurements were made without anti-scatter grid but no correction was made for attenuation by the protective layers above the detector. A 10mm x 10mm region of interest (ROI) positioned on the centre of each image was used for calculating the mean pixel value and the standard deviation of the pixel values. The relationship between mean pixel values and the detector entrance air kerma was determined to give the detector response curve of each system. These curves were subsequently used to invert the images from the DR system and linearise the images from the CR systems in a way such that the mean pixel value of each image will be equal to the air kerma used to acquire the image.

6.2.3 MTF measurement

When the system is undersampled, as in the case of digital systems, it is more common to use the presampling MTF (preMTF) which is the MTF of the system prior to the sampling process [12]. In this work, all the preMTF measurements were made using an edge method in a manner similar to that of Samei [1]. The edge test device comprised a 0.8mm thick rectangle (120mm x 60mm) of stainless steel with very sharp well-polished straight edges. For each measurement, the edge device was placed
in contact, centred on the surface of the detector (cassette in the CR system) and slightly angled (less than 2°) with respect to the rows (parallel to the chest wall edge) or columns (perpendicular to the chest wall edge) of the image matrix, as shown in Fig. 6.2. Five repeated measurements were made to estimate the standard error of the mean.

For the MTF calculation, the OBJ_IQ program was used [13,14]. The procedure used by the program for calculating MTF is as follow: from each linearised image, a 5cm x 5cm ROI centred on the region of the edge was first extracted for MTF analysis. Fig.6.3 shows an example of the MTF ROI used for MTF determination. The exact angle of the edge with respect to the detector array was determined by applying a first order fit to the edge transition data. The angle was then calculated as $\tan^{-1}(1/b)$, where $b$ is the gradient of the line image of the edge transition. The oversampled edge spread function (ESF) was created by re-projecting the image data along the direction of the edge line. This was done by plotting the value in each pixel versus the distance from the centre of the line edge. In the reprojection process, a sub-pixel binning factor of 0.1 was used, as this was found [1] to give an acceptable compromise between the
noise and resolution. The ESF was then smoothed with a median filter with a window size of 5 pixels. Fig.6.4 shows the ESF obtained using the Konica CR system with the CP-1M plate. The smoothed ESF was differentiated to give the line spread function (LSF), shown in Fig.6.5. The preMTF, in the direction across the edge line, was obtained by taking the modulus of the fast Fourier transform (FFT) of the LSF and normalising its value to 1 at zero spatial frequency.

Fig. 6.2. Experimental setup for MTF measurement. Note that unless stated all measurements were made with the compression paddle out of the beam.

Fig. 6.3. An edge image block extracted from an image of the edge device used for MTF analysis
6.2.4 Effect of X-ray spectrum on MTF measurements

In mammography the only useful part of the X-ray energy spectrum is the optimal energy. Therefore, the ultimate goal is to shape the spectrum as narrow as possible around this energy. In fact, the energy spectrum depends on several parameters such as anode material, filtration and maximum tube voltage [15]. The influence of applied tube voltage on the MTF measurement was investigated using the GE system. At a
fixed tube loading (mAs), the MTF of the system was measured for a range of selected tube voltages. The measurements were repeated using a fixed tube voltage (28kVp) with different mAs values. This measurement was also made using the Fuji and Konica CR systems.

The GE system has a selectable dual track anode, either molybdenum (Mo) or rhodium (Rh) with selectable filtration of Mo or Rh. The MTF of the system was measured using three target/filter combinations; Mo/Mo, Mo/Rh and Rh/Rh. The measurements were performed using the automatic exposure control (AEG) selected dose of each target/filter combination.

In clinical applications the radiation beam of a mammography system is attenuated by breast tissue. Therefore, the image performance of a mammography system should be evaluated using a beam quality as close as possible to that used in clinical applications. This can be achieved using an attenuator (added filter) placed in the beam. In the literature, evaluation of MTF of digital mammography systems was made with [1,10-18] and without [19-22] added filters in place. Different types of material with different thicknesses were used to filter the beam in the MTF measurement. These include aluminium [1,16,18], and PMMA [10,17] and recently an additional 2mm aluminium filter was recommended to be added [3]. The influence of added filtration on the accuracy of measuring MTF was investigated using the Konica CR system. The MTF of the system was measured with added filtration of a 2 mm thickness of aluminium foil and PMMA with a range of thicknesses from 20 to 55 mm. The MTF was also measured with no filtration added and the results were compared. The impact of the position of the added filter on the MTF measurement was also explored using the GE system. The MTF of the system was measured with a 45 mm PMMA mounted first at the tube-head and then halfway between the tube and the detector and the results compared.

6.2.5 Effect of clinical conditions on MTF measurements

In clinical diagnostic applications the breast is imaged with the anti-scatter grid and compression paddle in place. In the MTF measurement, however, the anti-scatter grid is always removed [16,20,22], as recommended in the IEC protocol [3]. Other publications [1,10,17-19] did not specify whether or not use the grid in their measurements. Most of the literature does not clarify whether or not the compression
paddle was in place during the MTF measurement. Also there is no recommendation in the IEC protocol regarding the paddle. In this work, the MTF of the GE system was measured with compression paddle in place. The paddle was placed at about 6cm above the detector surface. This measurement was then repeated without the compression paddle and the results compared. The MTF of the Giotto system was determined with and without the anti-scatter grid to investigate the impact of the grid on the MTF determination.

6.2.6 Effect of edge device on MTF measurements

The effect of the edge device on MTF measurement was previously studied in terms of edge device material, size, thickness uniformity [10] and transparency [23]. It was found [10,23] that the most important factors that might have an effect on the accuracy of MTF measurement are the thickness variation and the opacity of the edge foil. In this chapter, the effect of thickness of the edge device on the MTF measurement was evaluated using the Fuji CR system. Under the same exposure conditions MTF of the system was measured using two edge devices of different thicknesses. These are 100mm x 80mm x 2mm and 120mm x 60mm x 0.8mm stainless steel plates. Unless stated, all other MTF measurements in this chapter were made using the thinner edge device. The effect of location and orientation of the edge device relative to the detector array was also verified using the Konica CR system. The MTF of the system was measured with the edge device centrally placed and off-centre.

6.2.7 Directional dependence of MTF of CR systems

MTF of CR systems is mainly limited by light scattering in the phosphor layer of the CR powder plate during the laser reading stage [16,24]. Reducing the thickness of the phosphor screen may lead to reduce scattering of the laser light and hence improve the MTF of these systems [24]. This would, however, compromise the absorption efficiency of the plate. Therefore, the use of a needle plate leads to reduced light spread and therefore better image sharpness than the powder plate of equal thickness [25]. However, the MTF of CR systems depends on the direction in which it is measured [16,17,26,27]. In the scan direction, the MTF is determined by the laser scan speed and the luminance decay time, whereas, in the subscan direction it is limited by the lateral light scatter in the phosphor [16]. Therefore, this improvement...
in the MTF might not be achieved in the scan direction. In this chapter, the MTF of
the Konica CR system was measured using CP 1M and RP 7M plates but the same X-
ray set and laser reader to further investigate the directional dependence of the MTF
of CR systems.

6.3 Results and Discussions

6.3.1 Detector response and linearity

The linearity of the systems was verified by plotting the mean pixel values of the
uniform images as a function of the detector entrance air kerma used to obtain the
images. For the DR systems, the response was found to have a linear relationship with
a pixel value offset. The images from the CR systems had as expected non-linear
responses. Fig. 6.6 shows the detector response curve obtained for the GE DR system
with a linear function fit. Fig. 6.7 shows the non-linear response of pixel values to
incident air kerma for the images from the Fuji CR system with a logarithmic function
fit. Table 6.7 summarizes the fitted functions and their coefficients used to invert the
images from the DR system and linearise the non-linear images from the CR systems
for the MTF calculations.

![Graph](image.png)

**Fig. 6.6.** The linear response of the GE Essential system. The error bars indicate the 2 standard error of
the mean of the measurements.
Table 6.2. Fitted functions and coefficients used to linearise the images of the systems used here

<table>
<thead>
<tr>
<th>System</th>
<th>Fitted function</th>
<th>Offset (a)</th>
<th>Gradient (b)</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE Essential</td>
<td>Linear</td>
<td>0.0 ± 1.1E-12</td>
<td>8.40 ± 0.03</td>
<td>1.00</td>
</tr>
<tr>
<td>Giotto IMS Image MD</td>
<td>Linear</td>
<td>17.82 ± 8.35</td>
<td>7.19 ± 0.05</td>
<td>0.9999</td>
</tr>
<tr>
<td>Fuji Capsula CR</td>
<td>Logarithmic</td>
<td>-73.5 ± 1.1</td>
<td>111.9 ± 0.2</td>
<td>0.9999</td>
</tr>
<tr>
<td>Konica Regius 190 CP 1M</td>
<td>Logarithmic</td>
<td>12.614 ± 1.705</td>
<td>407.1 ± 0.5</td>
<td>0.9988</td>
</tr>
<tr>
<td>Konica Regius 190 RP 6M</td>
<td>Logarithmic</td>
<td>-68.64 ± 12.56</td>
<td>405.55 ± 0.93</td>
<td>0.9982</td>
</tr>
<tr>
<td>Konica Regius 190 RP 7M</td>
<td>Logarithmic</td>
<td>-72.39 ± 11.31</td>
<td>403.23 ± 0.82</td>
<td>0.9979</td>
</tr>
</tbody>
</table>

Fig. 6.7. The non-linear response of the Fuji CR system. The error bars indicate the 2 standard error of the mean of the measurements

6.3.2 MTF measurement

The preMTF of the DR and CR systems used in this work was measured and presented in Fig. 6.8. For validation, the MTF curves of the Konica CR system were compared to those obtained by other authors [16] and they are in good agreement as shown in Fig. 6.9. The system with similar plate (RP 6M) and reader (Regius 190) was used in both works. Results of the other systems can not be validated against the published study because of the variation in the models and versions.

The direct DR system (Giotto IMS) has a high preMTF compared to all other systems investigated. This trend could have been expected as this system uses direct
conversion a-Se technology which absorbs X-rays and converts them directly into electrical charge signal. The preMTF of the indirect DR system (GE Essential) is lower than for the direct system. The explanation of this is that the indirect system uses a phosphor layer in which the absorbed X-rays are converted into visible light which is then detected by a photodiode which constitutes the pixels of the detector. The light produced by the phosphor can, however, move laterally and scatter over several pixels of the detector limiting the effective resolution of the system. Moreover, this model of the GE system has been introduced with a large active field of view (24 cm x 30.7 cm) and with implementation of some changes in the flat panel design which might degrade its spatial resolution. The low preMTF of this model was also reported in a recently published study [28] in which it was concluded that although the MTF was much lower for the GE Essential than for the GE Senographe DS, the DQE for the Essential and the contrast-detail detection capability were significantly better. However, the MTF values reported in that study, using a beam quality recommended by the IEC protocol, i.e., 28 Mo/Mo and 2 mm of added aluminium filtration, were 0.85, 0.59, and 0.24 at 1, 2, and 4 mm⁻¹. These are in good agreement with the values presented here (0.84, 0.6, and 0.26 at 1, 2, and 4 mm⁻¹). Also the values presented here are comparable with the results of the internal image quality signature test (IQST) (0.6 and 0.25 at 2 and 4 mm⁻¹).

The preMTFs of the CR systems also suffer from light scatter in the phosphor layer and are lower than for the direct DR system. The Konica CR system has, however, a high preMTF, in the subscan direction, compared to the other CR system and the indirect DR system. This was achieved with the CP-IM plate which uses needle crystal technology where the lateral light scatter is significantly reduced. The preMTF of the two CR systems is much higher in the sub-scan direction than in the scan direction. The directional dependence of the preMTF of the Konica system is relatively higher than that of the Fuji system. The reasons for that will be discussed in detail in the last subsection of this section.

6.3.3 Effect of X-ray spectrum on MTF

Fig.6.10 shows the MTF of the GE system measured as a function of kVp. The measurements were made at three kVp values, 25, 29 and 34. The results are almost identical with very small differences. The preMTF measured at 25 kVp is slightly
lower than that at 29 and 34 kVp by about 1% and 1.5% for frequencies of 2 and 5 mm\(^{-1}\), respectively. It should be noted that the standard error of the mean of the MTF measurement is 0.6% and 1.9% for frequencies of 2 and 5 mm\(^{-1}\), respectively. This means that the MTF measurement has insignificant dependence on the X-ray energy. Fig.6.11 shows the MTF of the system measured at 29 kVp for three selected doses, 58, 117 and 233 μGy entrance air kerma. The results are almost identical with an error of about 0.5 and 1.5% for frequencies of 2 and 5 mm\(^{-1}\). Fig.6.12 shows the MTF of the Fuji CR system measured at 27 kVp Mo/Mo for three selected doses, 173, 351 and 707 μGy entrance air kerma. The results are almost identical which means that the dose has no influence on the MTF measurement. This behaviour was also observed for the Konica CR CP 1M as shown in Fig.6.13.

![MTF graph](image)

Fig.6. 8. Presampled MTF of the digital mammography systems used in this work. All the measurements were made with 2mm aluminium added filtration except for the Giotto system where 45mm PMMA was used, which presumably explains the large LFD in the MTF curve of this system.

The influence of the target/filter combination on the MTF measurement is illustrated in Fig.6.14. The preMTF of the GE system was measured for the available target/filter combinations, Mo/Mo, Mo/Rh and Rh/Rh. The results are very similar with minute differences observed in the higher frequencies. These differences are, however, within the standard errors of the mean of the MTF measurements. This means that the MTF
of a mammography system of this type appears to be independent of the clinically available target/filter used.

Fig. 6.9. Validation of MTF measurement against previous work done by Monnin et al (2007) [16]

Fig. 6.10. preMTF of the GE Essential system measured as a function of kVp and plotted up to the Nyquist frequency of the system. All measurements were made with a spectrum of Rh/Rh, 56 mAs and 45 mm PMMA added filtration. The error bars indicate the 2 standard error of the mean of the measurements.
Fig. 6.11. preMTF of the GE Essential system measured as a function of dose and plotted up to the Nyquist frequency of the system. All measurements were made with a spectrum of 29 kVp Rh/Rh and 45 mm PMMA added filtration. The error bars indicate 2 standard error of the mean of the measurements.

Fig. 6.12. preMTF of the Fuji CR system plotted as a function of dose. All measurements were made with a spectrum of 27 kVp Mo/Mo and 2 mm aluminium added filtration. The standard error of the mean of the measurements is very small to display (< ± 1% in average)
Fig. 6.13. preMTF of the Konica CR CP-1M plotted as a function of dose. All measurements were made with a spectrum of 28 kVp Mo/Mo and 2 mm aluminium added filtration. The 2 standard error of the mean of the measurements is <± 5% in average.

Fig.6.15 shows the MTF of the Konica CR system with the CP 1M cassette obtained with different situations of added filtration, 2 mm aluminium, PMMA of a range of thickness (20 mm – 55 mm) and no filtration added. Fig.6.16 shows the MTF measurements shown in Fig.6.15 enlarged at the lower range of spatial frequencies. The results show that the MTF obtained using 2mm aluminium is almost identical to that obtained with no added filtration. The MTF values obtained using PMMA are lower than that obtained using 2mm aluminium. They also show that the MTF values decrease as a function of thickness of PMMA filtration added. This was also observed using the RP 6M and RP 7M cassettes as shown in Figs.6.17 and 6.18, respectively. In a previous publication [10], it was found that the low frequency drop (LFD) increases with the thickness of PMMA filtration added. In the present work it was found that the MTF measurements decrease by 0.72 ± 0.12 %, 1.2 ± 0.2 %, 1.8 ± 0.3 % and 1.7 ± 0.6 % for every 5mm PMMA added filtration for the frequencies of 0.5, 1, 2 and 3 mm⁻¹, respectively. The use of PMMA to attenuate the beam at the tube-head introduces scattered radiation that distorts the MTF measured and therefore such use of PMMA should be avoided.
6.3.4 Effect of clinical conditions on MTF

The MTF of the GE system measured with and without compression paddle is shown in Fig.6.19. It can be seen that the LFD increases by $1.5 \pm 0.3 \%$ when the compression paddle is in place. This is due to the additional scattered radiation from the material of the paddle. To investigate the effect of the location of the added filtration on the MTF measurement, the premTF of the GE system was measured with 45 mm PMMA mounted at the tube-head. The measurement was repeated under the same conditions with the PMMA placed halfway between the tube and the detector and the results are shown in Fig.6.20. The MTF measurements were found to be strongly affected by the position of the PMMA. The LFD decreases by about $4 - 6 \pm 0.3 \%$ when the PMMA is placed on the tube-head.

According to the IEC protocol, the MTF should be measured with the anti-scatter grid removed. In this work, the premTF of the Giotto system was measured with and without the grid and the results are shown in Fig.6.21. While the LFD was not
affected, the MTF measurements were significantly overestimated by using the grid, notably at frequencies greater than 1 mm\(^{-1}\).

The IEC protocol states that the irradiated area of the detector surface should be 100 mm \(\times\) 100 mm. This can be done using additional collimation to cover the regions outside the irradiated area. This may, however, affect the final MTF measurements. Fig 6.22 shows the preMTF of the GE system measured with and without additional collimation. It can be seen that the well-collimated beam strongly reduces the LFD and increases the measured MTF at the higher spatial frequencies.

![Fig. 6.15. preMTF of the Konica CR system with CP-1M plate measured with different materials and thicknesses of added filtration. Measurement made with no filtration added is also included. The 2 standard error of the mean of the measurements is \(< \pm 5\%\) in average.](image-url)
Fig. 6.16. preMTF measurements shown in Fig. 6.15 enlarged at the lower range of spatial frequencies. The 2 standard error of the mean of the measurements is $< \pm 5\%$ in average.

Fig. 6.17. preMTF of the Konica CR system with RP-6M plate measured with different materials and thicknesses of added filtration. The 2 standard error of the mean of the measurements is $< \pm 5\%$ in average.
Fig. 6.18. preMTF of the Konica CR system with RP-7M plate measured with different materials and thicknesses of added filtration.

6.3.5 Effect of edge device on MTF

Fig. 6.23 shows the MTF of the Fuji CR system calculated using two test edge devices of different thicknesses; 2mm and 0.8 mm. The use of the two devices led to the same results. This indicates that, under conditions similar to those used here, a device with a thickness of 0.8 mm is suitable for MTF measurements for digital mammography with the advantage that no special heavy-duty mounting device is required.

Figs. 6.24 and 6.25 show the preMTF of the Konica CR system measured with the edge device centered on the detector surface and 2, 4 and 6 cm off-centre for the subscan and scan directions. It was verified by others [24] that moving the edge device by up to ± 2.5 cm off-centre would not affect the MTF measurements. In the present work it was found that moving the edge device up to ± 4 cm off-centre would not affect the measured MTF by more than 0.565 % ± 0.007 % at low frequencies in both directions.
Fig. 6.19. preMTF of the GE Essential system measured with and without compression paddle. All measurements were made with 45 mm PMMA added filtration. The error bars indicate the 2 standard error of the mean of the measurements is < ± 5% in average.

Fig. 6.20. preMTF of the GE Essential system measured using 45 mm PMMA added filtration placed on the tube and halfway between the tube and the detector. The error bars indicate the 2 standard error of the mean of the measurements is < ± 5% in average.
Fig. 6.21. preMTF of the Giotto system measured with and without ant-scatter grid. All measurements were made with 45 mm PMMA added filtration. The error bars indicate the 2 standard error of the mean of the MTF measurements.

It is known that for CR systems the sharpness strongly depends on the orientation of the edge relative to the direction of the movement of the laser beam in the scan direction [3]. Therefore, four edge images, rotating the test device over 90 degrees between each image, are required for MTF measurement of these systems [3]. However, by using an edge object of width 6cm and two sharp well-polished edges centered in the beam (so that both edges are 3 cm off-centre), a single image would be enough for each direction.

6.3.6 Directional dependence of MTF of CR systems

Fig.6.26 shows the preMTF of the Konica CR system measured in the subscan direction for CP-1M and RP-7M image plates using the same X-ray set and laser reader. Fig.6.27 shows the preMTF of these plates measured in the scan direction, as expected the MTF of the system was improved in the subscan direction, by using needle crystal technology. The improvement was about 13.5 ± 0.1 % and 42.1 ± 0.4 % at the spatial frequency of 2 and 11.43 mm⁻¹ (the Nyquist frequency of the system), respectively. In the laser scan direction, on the other hand, the MTF of the two image
plates were very similar. This indicates the strong dependence of the MTF of CR systems on direction in which it is measured. The resolution property of these systems is determined by the light scatter in the phosphor in the subscan direction, whereas, in the scan direction it is defined by the laser scan speed and the luminance decay time. This means that reducing light spread in the phosphor plate, by using the needle photostimulable plate, would improve the spatial resolution only in the plate direction. Therefore, further improvements in the CR laser reader appear necessary in order to improve the MTF in the scan direction. This could be done by making a compromise between the laser scan speed and the luminance decay time. Nevertheless, most of the current CR readers do operate logarithmically in the scan direction and therefore, a true MTF of the system can only be defined in the subscan direction [29].

Fig. 6.22. preMTF of the GE Essential system measured with and without additional collimation. All measurements were made with 45 mm PMMA added filtration. The error bars indicate the 2 standard error of the mean of the MTF measurements.
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Fig. 6.23. preMTF of the Fuji CR system measured using two edge devices with different thicknesses. The error bars indicate the 2 standard error of the mean of the MTF measurements.

Fig. 6.24. preMTF of the Konica CR system with CP-1M plate, measured in the subscan direction with the edge device on and off centre of the detector surface. The 2 standard error of the mean of the measurements is < ± 5% in average.
Fig. 6.25. preMTF of the Konica CR system with CP-1M plate, measured in the scan direction with the edge device on and off centre of the detector surface. The 2 standard error of the mean of the measurements is $\leq \pm 5\%$ in average.

Fig. 6.26. preMTF of the Konica CR system measured in the subscan direction for cassettes CP-1M and RP-7M. The error bars indicate the 2 standard error of the mean of the MTF measurements.
Fig. 6.27. preMTF of the Konica CR system measured in the scan direction for cassettes CP-1M and RP-7M. The error bars indicate the 2 standard error of the mean of the MTF measurements.

6.4 Conclusions

The determination of the spatial resolution of an imaging detector with the edge method may in principle be affected by several factors e.g. tube voltage and current, target/filter combination, type, position and thickness of added filtration, presence of compression paddle and anti-scatter grid, the radiation field size and the radiation scattered from the edge test device. In this chapter the influence of most of these factors on the MTF measurements was investigated for digital mammography systems. Overall, there was little or no variation in the MTF measurements as a function of kVp, dose and target/filter combination.

The study has shown that the thickness of the PMMA added filtration has a pronounced effect on the MTF results especially at the low frequencies. However, the use of 2mm aluminium led to a LFD that was essentially the same as if no filtration had been added. Therefore, to obtain an MTF with an accurate LFD and to protect the detector from radiation damage the X-ray beam should be attenuated using a 2mm thickness of aluminium mounted at the tube-head.

It was found that the use of the anti-scatter grid affects the measurements in a way that leads to the true detector MTF being overestimated in the mid-high frequency
range. This suggests that the MTF should be determined without the ant-scatter grid.
The use of a well-collimated beam was found to compensate for the absence of the grid by reducing excessive scattered radiation. Although the use of the compression paddle was found to have a very small effect on the MTF results, it is recommended to remove it during the measurement.

The results have shown that the determined MTF would not be affected by moving the edge device up to ±4 cm off-centre. This would suggest that for CR systems, using an edge device of 6 cm wide with two sharp well-polished edges centered in the beam, two images could be acquired instead of four images which would otherwise be required due to the strong dependence of the MTF on the orientation of the edge device. Finally, using a needle imaging plate with a CR system improves the MTF only in the subscan direction of the system. In the scan direction the MTF remains the same as if a conventional phosphor plate is used. Therefore, it may be concluded that improvements in the CR laser reader are needed to improve the MTF of such a system in the scan direction.

References


7. ICRU report 41, Modulation transfer function of screen-film systems, International commission on radiation units and measurements, USA, 1986


Chapter 7

Predicting Contrast Detail Performance from Objective Measurements*

7.1 Introduction

The performance of a digital mammography system can be quantified and compared in a number of different ways. European and UK guidelines define procedures for measuring image quality in terms of contrast detail detectability, based on readings of images of the CDMAM test object by human observers [1,2]. This method is, however, time-consuming and has large inter- and intra-observer errors. As described in chapter 5 a possible solution to these problems is software, which allows automatic reading of CDMAM images. The automated measurements can be then used to predict the threshold contrast for a typical observer. This approach, however, suffers from a limitation that using an automatic method produces results that are different from those of human observers. In chapter 5, a method of predicting the threshold contrast observed by a typical human observer from automatic reading was reported. However, the automatic method requires a large number of images to reduce statistical errors. Therefore, an alternative approach would be to predict threshold contrast from objective measurements, such as MTF and DQE.

Measurements of MTF, NNPS, NEQ and DQE are well established for assessing detector performance [3]. The physical basis of these metrics and their relevance to the subjective measurements of radiological images has been extensively studied by Wagner and Brown [4]. They proved that the detection of details in X-ray images by a human observer can be modelled by statistical decision theory. This was then put into

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practice by Workman and Cowen [5] for computed radiography (CR) systems, Marshall [6] for a-Se direct radiography (DR) mammography system and more recently by Rivetti and co-workers [7] for one CR system and two DR systems. However, more research is required before such an approach could replace the well established quality control procedures specified by the European Guidelines for quality control in digital mammography. This study is an attempt to further explore the feasibility of this approach using multi CR and DR mammography systems.

It is well known that the detector measurements suffer from two main difficulties [8]. The first is that these measurements do not include some other factors that affect the overall system performance such as the effect of the X-ray spectrum on the contrast, and the effect of scatter on contrast degradation. The second difficulty is that the different systems use a range of radiation doses. Thus, when comparing systems, allowance must be made for the different doses used. In this work, an attempt to overcome these limitations was made. This was done by analysing the contrast detail performance of a wide range of DR and CR mammography systems across a range of doses and presenting the results as the dose required to achieve the standards in the European guidelines at different detail sizes. MTF, NNPS, NEQ and DQE were also obtained for the detectors used in these systems following the IEC protocol [9]. The contrast detail measurements of these systems were then modelled using a simple signal-matched noise-integration model [4]. In general, the aims of this chapter are

- to evaluate the performance of a wide range of the currently available digital mammography systems by comparing their detector performance
- to consider whether the performance measured by contrast detail measurement is consistent with that expected from the results on detector performance
- to further investigate the feasibility of using the objective measurements to predict the contrast detail measurements

7.2 Materials and Methods

7.2.1 Systems Tested and Image Acquisition

The study was performed using a wide range of digital mammography systems. Table 7.1 illustrates the detector and image specifications for these systems. Each
measurement was made at a hospital or centre where the unit has been installed. All images acquired during the measurement were saved as unprocessed files in DICOM format. These are then transferred to a personal computer for later analysis.

Table 7.1. Physical characteristics of the Systems used in the study

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>Model</th>
<th>Detector</th>
<th>Pixel size (µm)</th>
<th>Mammography X-ray set</th>
<th>Image plate</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hologic</td>
<td>Selenia</td>
<td>a-Se direct DR</td>
<td>70</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>GE Medical Systems</td>
<td>Essential</td>
<td>a-Si indirect DR</td>
<td>94</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Giotto IMS</td>
<td>Image MD</td>
<td>a-Se direct DR</td>
<td>80</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Siemens</td>
<td>Inspiration</td>
<td>a-Se direct DR</td>
<td>85</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Sectra</td>
<td>MDM-L30</td>
<td>Photon counter</td>
<td>50</td>
<td>GE DMR</td>
<td>HR-BD</td>
</tr>
<tr>
<td>Fuji Photo Film</td>
<td>FCR Capsula</td>
<td>Single-side CR</td>
<td>50</td>
<td>GE DMR+</td>
<td>HR-BD</td>
</tr>
<tr>
<td>Agfa CR</td>
<td>CR85-X</td>
<td>Single-side CR</td>
<td>50</td>
<td>MM3.0</td>
<td>-</td>
</tr>
<tr>
<td>Kodak</td>
<td>CR850</td>
<td>Single-side CR</td>
<td>50</td>
<td>EHR-M2</td>
<td>-</td>
</tr>
<tr>
<td>Konica Minolta</td>
<td>Regius 190</td>
<td>Single-side CR</td>
<td>43.75</td>
<td>GE DMR</td>
<td>RP-6M</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>RP-7M</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>CP-1M</td>
</tr>
</tbody>
</table>

7.2.2 Detector response and linearity

In order to measure the MTF and NPS of a detector the detector has to be linear and shift invariant [10]. The detector response curve of each system was obtained using the method described in the previous chapter. These curves were subsequently used to invert the images from the DR system and linearise the images from the CR systems in a way such that the mean pixel value of each image will be equal to the air kerma used to acquire the image.

The difficulties of measuring the MTF of the non shift-invariant digital detectors was discussed in details in chapters 2 and 6. NPS measurements will vary depending on spatial location, due to the shift-variant property of these detectors. However, in this work shift-invariant detectors are assumed [11,12]. All ROIs used for the NPS measurements for all systems were taken from the same location of multiple images to allow for this assumption. The same approach was adopted by Vedantham and co-workers [12].

7.2.3 Physical measurements

1. MTF

The preMTF of each system was measured using the optimised method concluded in the previous chapter. The dose used to acquire the MTF and NPS images was measured using methods as close as possible to that described by the IEC protocol [9].
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For each system, flat field images were obtained using a beam quality adjusted by placing a uniform 2mm thick aluminium filter at the tube housing, at multiple exposure levels. Table 7.2 illustrates the radiation quality used for each system. For each acquisition, the entrance air kerma on the detector was measured using a calibrated ion chamber, as described in the previous chapter. For each exposure, five flat field images were acquired in order to calculate the average air kerma incident on the detector surface.

Table 7.2. Radiation quality and number of photons incident on the detector for all systems

<table>
<thead>
<tr>
<th>System</th>
<th>kVp</th>
<th>Target/Filter</th>
<th>HVL (mm Al)</th>
<th>Quanta/mm²/μGy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hologic Selenia</td>
<td>28</td>
<td>W/Rh</td>
<td>0.75</td>
<td>5975</td>
</tr>
<tr>
<td>GE Essential</td>
<td>28</td>
<td>Rh/Rh</td>
<td>0.74</td>
<td>5944</td>
</tr>
<tr>
<td>Giotto IMS</td>
<td>30</td>
<td>W/Rh</td>
<td>0.80</td>
<td>6430</td>
</tr>
<tr>
<td>Siemens Inspiration</td>
<td>28</td>
<td>W/Rh</td>
<td>0.75</td>
<td>5975</td>
</tr>
<tr>
<td>Sectra MDM-L30</td>
<td>29</td>
<td>W/Al</td>
<td>0.88</td>
<td>6925</td>
</tr>
<tr>
<td>Fuji Capsula</td>
<td>27</td>
<td>Mo/Mo</td>
<td>0.56</td>
<td>4863</td>
</tr>
<tr>
<td>Agfa CR85-X</td>
<td>30</td>
<td>Mo/Rh</td>
<td>0.68</td>
<td>5730</td>
</tr>
<tr>
<td>Kodak CR850</td>
<td>27</td>
<td>Mo/Rh</td>
<td>0.64</td>
<td>5460</td>
</tr>
<tr>
<td>Konica Regius 190</td>
<td>28</td>
<td>Mo/Mo</td>
<td>0.58</td>
<td>5007</td>
</tr>
</tbody>
</table>

II. NPS

The NPS was calculated using the OBJ_IQ program [13,14]. The procedure used by the program for calculating NPS is as follow: a 1024 × 1024 central region of interest (ROI) was first extracted from the linearised CR and DR flat field image. Non-overlapping 64 sub-ROIs, each 128 × 128 pixels in size were then taken from the large ROI [6]. This size was selected because it was found [6,16] to be the smallest ROI size that could be used without appreciably underestimating the NPS curve near zero frequency. At this stage the data in these sub-ROIs vary from pixel to pixel due to the non-uniformity in the X-ray field. This would corrupt the noise spectrum and produce incorrect values along the axes [12]. However, methods for removing the background trends caused by the non-uniformities have been described in the literature [12,15,16]. In this work, the data within each sub-ROI was corrected for the presence of the background trends by first fitting a two-dimensional second-order polynomial and then subtracting the fit from the data [6].

For each selected exposure, the ensemble average of the squares of the magnitude of the 5 Fourier transformed ROIs was obtained as shown in Eq. 2.9 to give the 2-D NPS of the detector used in each system. This was then multiplied by the squared pixel size
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and divided by the number of pixels, in the two dimensions, to express the NPS in units of area (mm²), as per Eq.2.9. The 1-D NPS was estimated by averaging five lines on either side of both \( u \) and \( v \) axes (excluding the axes) within the 2-D NPS array. The non-uniformities correction method used here is successful in suppressing the background trends but not completely eliminate them [17]. Therefore, the data values directly on the axes were excluded while estimating the 1-D NPS from the 2-D NPS. Finally, the 1-D NPS was normalised to the squared mean pixel value of the linearised subtracted ROI to give the normalised NPS (NNPS). This was done to eliminate direct effect of signal variations between the images or ROIs used for the NPS [18].

III. NEQ and DQE

The NEQ was computed for each system using Eq. 2.11. The DQE curves were then obtained using Eq. 2.10. For the DQE calculation, the precise number of photons incident on the detector need to be determined. For each beam quality the entrance air kerma was measured as described earlier. This was then used to obtain the number of quanta using conversion factors tabulated in the IEC protocol [9] and shown in Table 7.2. For the beam qualities not presented in the IEC protocol, the corresponding parameters (e.g. photon flux) were calculated by the programme provided by The Institute of Physics and Engineering in Medicine [19] as part of a spectral catalog on CD-ROM. In general, the software produced results higher than that presented in the protocol by less than 1%. For consistency, this variation was considered in the calculated data.

7.2.4 Contrast-detail measurements

The threshold gold thickness for each diameter was determined using human and automatic readings of the CDMAM images at the different levels of exposure, as described in chapter 5. The CDMAM test object was first exposed at the exposure factors selected by the system's AEC control. A set of at least 8 images were obtained. These exposures were repeated using the beam quality selected by the AEC but at lower and higher mAs values. For each set of exposure factors the mean glandular dose (MGD) of a breast equivalent to the test phantom was calculated using Eq. 5.1 in chapter 5.
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7.2.5 Comparing the performance of systems

As mentioned earlier, the detector physical measurements do not include some other factors that affect the overall system performance such as the effect of X-ray spectrum and scatter on the image contrast. Another complication is that the different systems use a range of X-ray doses. Therefore, in order to make a comparison between different systems allowance must be made for the different doses used. In this work, the contrast detail performance of different systems was determined across a range of doses. The results were then presented as the dose required to achieve the standards specified in the European guidelines at different detail sizes. This was made by fitting curves to the datasets for each system as per Eq. 7.1,

\[ T = k \text{MGD}^n \]  

where \( T \) is the determined threshold gold thickness, \( k \) and \( n \) are constants to be fitted, and \( \text{MGD} \) is the mean glandular dose for a breast of thickness equivalent to 60mm of PMMA. The value of \( n \) was constrained to be the same for a set of curves for a given system. If quantum noise is dominant \( n \) is expected to have a value of approximately 0.5 [8]. The fitted curves were used to calculate the MGD required at the limits stated by the European guidelines for each detail diameter. At diameters where these are not specified the limits were interpolated. Thus each system was characterised by the dose required to reach minimum and achievable image quality standards across a range of detail sizes.

7.2.6 Contrast-detail model

In the literature a number of psychophysical models have been applied to determine theoretically contrast detail detectability for imaging systems [20]. The two well known models are the Rose model [21], and the signal-matched noise-integration model [4]. The relationship between the two models is demonstrated at length elsewhere [22]. However, it was found [6], with some assumptions, that the two models led to the same agreement between the measured and the modelled contrast detail measurements. In this work the contrast detail detectability was modelled using the signal-matched noise-integration model. The model predicts that for detection to occur, the displayed signal-to-noise ratio (SNR) must be equal to some value, \( k \), known as the subjective threshold criterion, held by the observer [20,23]. This means
CHAPTER 7. PREDICTING CONTRAST DETAIL DETECTIBILITY

that for the task of detecting a circular object of area $A$, the threshold contrast $C_T$ of this task can be predicted as,

$$C_T(A) = \frac{k}{[SNR_\theta(A)A]^{0.5}}$$  \[7.2\]

where SNR is the displayed signal to noise ratio which is given by,

$$SNR^2 = (X_e . R . DQE(X_e))$$  \[7.3\]

where $X_e$ is the detector entrance air kerma ($\mu$Gy), $R$ is the conversion factor giving the number of X-ray quanta/mm$^2$ per unit exposure ($\mu$Gy), shown in Table 7.2, $DQE(X_e)$ is the DQE at zero frequency calculated at the detector entrance exposure level $X_e$. $\theta_s(A)$ is the SNR degradation factor due to unsharpness given by,

$$\theta_s(A) = A/(NEA_e . a_{dp} + A)$$  \[7.4\]

where $NEA_e$ is the noise equivalent aperture of the imaging system defined as,

$$NEA_e = (2\pi \int_0^{LR} MTF^2(f).f.df)^{-1}$$  \[7.5\]

where $LR$ is the maximum spatial frequency at which the system response is non-negligible. $a_{dp}$ is the unsharpness arising from display/perception, which is assumed to be negligible due to the fact that the observers were able to optimize their viewing conditions. A similar assumption was made by Marshall [6] for a-Se Hologic Selenia system and Workman and Cowen [5] for CR systems.

Given knowledge of the physical measurements and using the appropriate values of $X_e$ and $R$ for the detector systems it was possible to fit theoretical curves to the experimental contrast detail results, for details ranging in size from 1 mm down to 0.08 mm. The value of $k$ was adjusted to optimize the fit to all the curves. Because the observer's level of detectibility is usually taken to be 50%, $k$ is usually found to take on a value between 2 and 5 [23]. In previous works values of 4 [5], 3.8 [24] and 2.5 [6] were assumed for $k$. Other authors have used values of $k$ of 4.5, 5 and 6 for Fuji
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CR, GE and Giotto IMS DR systems, respectively [7]. In this work a k value of 2.6 was adopted, for all systems, which gave the best fit to the measured data.

The value of threshold contrast, $C_T(A)$, corresponding to each detail diameter in the CDMAM found by the model was used to calculate the threshold gold thickness as described in the European guidelines after allowing for an estimated contrast loss due to scatter. For low radiation contrast the primary contrast will be degraded by scattered radiation by a factor [23] of $(1+S/P)^{-1}$, where $S/P$ is the scatter – to – primary ratio. For conditions and situations similar to those used here, it was found that the primary contrast would be reduced due to scatter by a factor of 0.86 [23]. A similar value was also reported in a more recent publication [25].

7.3 Results and Discussion

7.3.1 Detector response and linearity

The linearity of the systems was verified by plotting the mean pixel values of the uniform images as a function of the detector entrance air kerma used to obtain the images. As discussed earlier, all the DR systems used here were found to have a linear response with a pixel value offset. The images from the CR systems had non-linear responses. Table 7.3 summarizes the fitted functions and their coefficients used to invert the images from the DR system and linearise the non-linear images from the CR systems for the MTF and NPS measurements. For the photon counting system, the detector response could not be measured as described in the NHSBSP protocol [2]. This is partly because an enclosed collimator prevents positioning of attenuating material close to the X-ray tube [26].

<table>
<thead>
<tr>
<th>System</th>
<th>Fitted function</th>
<th>Offset ($a$)</th>
<th>Gradient ($b$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hologic Selenia</td>
<td>Linear</td>
<td>53.3± 19.2</td>
<td>3.91± 0.03</td>
</tr>
<tr>
<td>GE Essential</td>
<td>Linear</td>
<td>0.0 ± 0.0</td>
<td>8.4 ± 0.03</td>
</tr>
<tr>
<td>Giotto IMS</td>
<td>Linear</td>
<td>17.8 ± 8.0</td>
<td>7.19 ± 0.05</td>
</tr>
<tr>
<td>Siemens Inspiration</td>
<td>Linear</td>
<td>53.5± 21.8</td>
<td>3.35± 0.02</td>
</tr>
<tr>
<td>Fuji Capsula CR</td>
<td>Logarithmic</td>
<td>-83.8 ± 12.7</td>
<td>111.1 ± 0.6</td>
</tr>
<tr>
<td>Agfa CR85-X</td>
<td>Power</td>
<td>Power to 0.50±0.09</td>
<td>77.10±0.10</td>
</tr>
<tr>
<td>Kodak CR850</td>
<td>Logarithmic</td>
<td>2993± 476</td>
<td>-410.1±0.75</td>
</tr>
<tr>
<td>Konica Regius 190 CP 1M</td>
<td>Logarithmic</td>
<td>-103 ± 14.0</td>
<td>425.1 ± 0.5</td>
</tr>
<tr>
<td>Konica Regius 190 RP 6M</td>
<td>Logarithmic</td>
<td>-144 ± 26.0</td>
<td>423.0 ± 1.0</td>
</tr>
<tr>
<td>Konica Regius 190 RP 7M</td>
<td>Logarithmic</td>
<td>-150.8 ± 24.0</td>
<td>421.9 ± 0.9</td>
</tr>
</tbody>
</table>
7.3.2 Physical measurements

I. MTF

The preMTF of the DR and CR systems used in this work was measured and plotted in Fig. 7.1. The validation of this measurement was made against previous work done by Monnin et al (2007) [27] with good agreement as reported in chapter 6. The superiority of the MTF of the DR systems over the CR systems and the deteriorated MTF of the Essential GE model were discussed in detail in the previous chapter. For the Sectra system, the preMTF measured in the slit scan direction is lower than that measured in the array direction, for spatial frequencies higher than 1 mm\(^{-1}\). While in the array direction the spatial resolution of this system is mainly determined by the pixel size, it is on the scan direction that it depends on the slit width of the pre-collimator and the scan movement [27]. However, the system has the highest preMTF of all the systems, in the array direction.

II. NPS

For all systems tested here, a variance image was initially obtained, using one of the NPS uniform images, and assessed for any detector artefacts that may be present. This
was made before calculating NPS as recommended by Marshall [6]. Fig. 7.2 shows an example of image variance of a healthy detector. This image shows a uniform variance across the chest wall region but increase as we move further away from the chest wall side to the nipple side, due to the heel effect. In this image there are no severe detector artefacts present. In contrast, Fig. 7.3 shows an example of a failed variance image from an un-healthy detector. Quite pronounced artefacts and some defective pixels are clearly demonstrated.

After the uniformity of the variance image had been verified, the two-dimensional NPS was computed. The 2-D NPS images are very important for further visual inspection of pixel value uniformity and presence of the significant artefacts [16]. For each system the 2-D NPS image was formed and assessed for uniformity. Figs. 7.4(a), 7.4(b), 7.4(c) and 7.4 (d) show typical examples of the 2-D NPS for a-Se Siemens Inspiration, a-Si GE Essential and Konica CR CP 1M acquired at 114, 89 and 183 and 90 μGy, respectively. For the direct DR system, the NPS is relatively uniform in all directions with a significant increase along the axes. The indirect DR system, on the other hand, exhibits a remarkable sharp drop in the NPS at high frequencies. Similar behaviours were observed for the other DR systems used in this work (results are not shown). The CR system showed elevated noise in the horizontal direction, presumably due to uncorrected heel effect. It also showed pronounced marks in the NPS of the high exposure image (183 μGy, Fig. 7.4, c), possibly due to slight system noise patterns. Generally, these marks are not seen in the low exposure NPS image (90 μGy, Fig. 7.4, d) since the ratio of X-ray quantum noise and luminescence to system noise increase as dose decreases [16,28]. Similar trends were also demonstrated (results are not shown) for the other CR systems used here.

The 2-D NPS data were used to estimate the 1-D NNPS required for the DQE analysis. The 1-D NNPS of the digital mammography systems were calculated at the AEC selected exposure level corresponding to each system. This exposure level was selected as that used when the system is operated for the intended use in clinical application. This is called the "typical" exposure which can be determined using the AEC control for a typical compressed breast, and usually given by the manufacturer. Two additional exposure levels were then chosen that were approximate multiples of...
2 higher or lower (i.e. double and half) than the typical level, as specified in the IEC protocol [9]. Table 7.4 illustrates the typical (normal) dose used for each system.

Fig. 7.2 Example of a variance image obtained from a healthy detector. Black indicates low variance and white indicates high variance.

Fig. 7.3 Example of a variance image obtained from a non-healthy detector. Quite significant artefacts and some defective pixels are also shown.
Fig. 7.4 Measured 2-D NPS for (a) direct DR at 69 μGy, (b) indirect DR at 50 μGy, (c) and (d) CR system at 183 and 90 μGy, respectively. The contrast scales were adjusted independently for each image to best display the NPS structure. The intersection of the axes has much higher value of NPS and hence this point was masked for display purposes.

The half and double doses are also shown for some systems. These systems were taken as an example of each technology of digital mammography. The behaviour of the NNPS and DQE of the different technologies of digital mammography will be studied and compared, using these selected systems. Fig. 7.5 shows the 1-D horizontal NNPS of the digital mammography systems for the typical dose level of each system. The standard error of the mean calculated for the NNPS measurements was approximately ± 3% at all frequencies.
Table 7.4. Normal dose used for the digital mammography systems

<table>
<thead>
<tr>
<th>System</th>
<th>Normal dose (μGy)</th>
<th>Half dose (μGy)</th>
<th>Double dose (μGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hologic Selenia</td>
<td>114±2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>GE Essential</td>
<td>50.4±1.0</td>
<td>24.6±0.06</td>
<td>96.3±1.0</td>
</tr>
<tr>
<td>Giotto IMS</td>
<td>171±3</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Siemens Inspiration</td>
<td>68.9±0.4</td>
<td>33.5±0.4</td>
<td>138.3±0.8</td>
</tr>
<tr>
<td>Sectra MDM-L30</td>
<td>46.3±0.9</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fuji Capsula CR</td>
<td>85.8±1.7</td>
<td>44.0±0.9</td>
<td>173.3±3.5</td>
</tr>
<tr>
<td>Agfa CR85-X</td>
<td>132±3</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Kodak CR850</td>
<td>86.9±1.7</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Konica Regius 190 CP</td>
<td>89.9±1.8</td>
<td>44.0±0.9</td>
<td>183.1±3.7</td>
</tr>
</tbody>
</table>

Fig. 7.5 Measured 1-D horizontal NNPS of the digital mammography systems at the corresponding typical dose level. For the CR systems the results are obtained from the plate direction (subscan) and for the Sectra the data from the array direction were used.

The indirect DR system has the lowest high-frequency NNPS. Its NNPS curve is, however, not constant over approximately all frequencies, possibly due to the low MTF of this version of the GE system. For direct DR systems (including the photon counter) the NNPS curves are relatively uniform over a large spatial frequency band. This can be attributed to their MTF which remains high, up to the Nyquist frequency. While the photon counter system has approximately the highest higher-frequency NNPS, the curve is quite constant over all frequencies. This is probably because of fewer conversion steps from X-rays to the electronic signal and the electronic
threshold method associated with photon counting [27]. However, the NNPS of this system was measured at the lowest dose used in this study.

The CR systems have in general high low-frequency NNPS which rapidly falls off as the frequency increases, although it is less pronounced for the Konica system. This behaviour is presumably related to the uncorrected heel effect and the so-called “mammogram quantum mottle” which increases as the X-ray absorption rate decreases. The reduction in the absorption efficiency would be remarkable in the CR plate with a very thin phosphor layer. In fact, there is a trade-off between the X-ray efficiency and the spatial resolution of CR systems. The thinner the phosphor layer of the CR plate, the higher is the spatial resolution but the lower is the quantum efficiency of the system. This was the case in all conventional powder CR plates used here in which the phosphor thickness must be kept thin enough to reduce lateral light spread and improve the spatial resolution of the system. This, however, leads to a decreased quantum efficiency and an increased quantum mottle. A possible compromise in this debate is using the so-called “needle crystal technology”.

The use of a needle plate leads to limit lateral light scatter and therefore improve the image spatial resolution while maintaining a reasonable phosphor thickness. As this technology was employed in the CP-1M plate of the Konica CR used here, the NNPS of the system does not show a significant increase at low frequency, which is more obvious, for the other CR systems. To further investigate how this approach would improve the NNPS at the low frequency, NNPS of the Konica CR system was computed for the needle CP-1M plate and plotted together with that for the powder RP-6M and RP-7M plates, as shown in Fig. 7.6. For appropriate comparison the measurements were made using the same X-ray set and CR reader for the typical exposure level with the same radiation quantity.

By using the needle crystal technology, the CP-1M plate exhibits lower NNPS at low frequency than those from the other two plates. It is of interest to note that CP-1M plate has also shown much higher MTF than that obtained using the RP-7M plate, as shown in Fig 6.26 in chapter 6, which in turn has a better MTF than that of RP-6M plate (results are not shown).
The NNPS curves were plotted in the horizontal, vertical and diagonal directions at the three levels of dose for the Siemens, GE and Konica CP-IM systems, as shown in Figs. 7.7, 7.8 and 7.9, respectively. It appears, for all systems, that the NNPS is strongly exposure dependant. It decreases with increasing exposure as the relative quantum noise increases with exposure. For each dose level investigated, the NNPS of the two DR units is relatively flat in all directions. It has also almost similar magnitude in all directions indicating excellent symmetry in all the frequency range. This good symmetry was clearly observed in the 2-D NPS image (Fig. 7.4, a). The NNPS curve of the direct DR system drops slowly with frequency, due to aliasing of noise from beyond the Nyquist frequency into lower frequency. The indirect DR system, in contrary, shows a significant sharp drop in the NNPS curve at high frequencies. This was clearly displayed in the 2-D NPS image of this system (Fig. 7.4, b).
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Fig. 7.7 The NNPS of the Siemens system calculated at all directions for the three dose levels.

Fig. 7.8 The NNPS of the GE system calculated at all directions for the three dose levels.
Fig. 7.9 the NNPS of the Konica CP-1M system calculated, at horizontal (u), vertical (v) and 45°
direction with respect to the image matrix, for the three dose levels

The NNPS results of the CR system display a significant asymmetry between the scan
and subscan directions. This is expected for the CR systems due to differences in
signal processing in the two directions [27]. This phenomenon was observed in the 2-
D NPS image of the system (Fig. 7.4 c and d).

It was assumed [29] that the noise of a detector would comprise three components;
electronic noise independent of exposure, quantum noise proportional to exposure and
structural (fixed pattern) noise which is proportional to the square of exposure. These
three noise sources have a relationship described by,

$$\sigma = \sqrt{k_e + k_q X + k_s X^2}$$  \[7.6\]

Where $\sigma$ is the standard deviation in pixel values of a uniform image, $X$ is the incident
air kerma level and $k_e$, $k_q$ and $k_s$ are the coefficients determining the amount of
electronic, quantum and structural noise, respectively. $\sigma$ was obtained and plotted
against dose for the three mammographic units examined. Fig. 7.10 shows the noise
components of the Siemens and GE systems. The results of the Konica CR system are
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shown in Fig. 7.11. Non-linear regression was fitted to all curves, to determine the different noise components.

Fig. 7.10 Variance in pixel values versus air kerma for the GE and Siemens systems. The equations, including the fitting coefficients are also shown. The 2 sem in is very small to display (<0.2% in all measurements)

Fig. 7.11 Variance measurements for the Konica CR CP-1M system. The equations, including the fitting coefficients are also shown. The 2 sem in is very small to display (<0.2% in all measurements)
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It can be seen that the electronic noise is appreciably lower on both the DR systems than on the CR system. The quantum and structural noise are quite similar for both DR systems. The structural noise is noticeably higher on the CR system than on the two DR systems. This would explain the presence of the streaks shown in the 2-D NNPS image of the CR system (Fig. 7.4, c). The negative sign in the noise equation of the CR system means that the ratio of X-ray quantum noise to system noise decreases as dose increases. As a result the streaks are less significant in the 2-D NPS of the low exposure image (Fig 7.4, d).

All other direct DR and CR systems used here have displayed (results are not shown) NPS behaviour similar to those observed for the Siemens and Konica CR systems, respectively. However, since the NPS measurements were made using different radiation qualities, it is hard to directly make a comparison between the different digital mammography technologies.

**III. NEQ and DQE**

Fig. 7.12 shows the NEQ, in the horizontal direction with respect to the image matrix, of the Siemens, GE, Konica and Sectra systems, obtained at the reference dose of each system. The dependence of NEQ on dose was investigated and plotted in Fig. 7.13 for the GE and Konica CR systems. For better display these systems were selected as representatives of the other DR and CR systems as they have quite similar levels of dose.

It can be observed that the NEQ is strongly dose dependent for both systems. It increases as the dose increases. This dependence is, however, more pronounced for the DR system than that for CR system. At half dose they have quite similar magnitude of NEQ for a wide range of spatial frequencies. As the dose is doubled the DR system shows a significant increase in the values of NEQ more than that for the CR system. The NEQ values increase by 85% and 62% from half dose to double dose at spatial frequency of 3.9 mm⁻¹, for the DR and CR systems, respectively. This was also observed (results are not shown) for the all other DR and CR systems. This performance of the NEQ of both systems is based on that for the CR plate the ratio of quantum noise to system noise decreasing as dose increases. The DR detector, on the other hand, appears to have quantum limited noise at the higher dose.
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Fig. 7.12 the NEQ ($u$) of the GE, Siemens, Sectra and Konica CR systems for the corresponding reference level of dose.

Fig. 7.13 the NEQs of the GE and Konica CR CP-1M systems for the three levels of dose investigated.
Nonetheless, due to this high dependence of the NEQ on dose, it is hard to directly compare the NEQ values of different DR systems calculated for different levels of dose. Therefore and for best comparison, it is better to use the DQE which is dose independent for digital DR detectors especially at doses high enough for electronic noise to be neglected and for the DQE to reach a plateau. Fig. 7.14 shows the DQE of the Siemens DR system calculated for the diagonal direction for a range of levels of dose. The standard error of the mean of the DQE measurements was approximately ± 6%. The uncertainty of the DQE values is actually expected to be less than ± 10%, as stated in the IEC protocol [9].

![Graph showing DQE vs. Spatial frequency for different dose levels](image)

Fig. 7.14 DQE($u,v$) of the Siemens DR system at five dose levels. The error bars indicate the 2 sem of the measurement.

It can be seen that the DQE curves increase as dose increases till they reach a plateau. The DQE drop at low dose is possibly due to the effect of the electronic noise which becomes innegligible relative to the useful signal. This was observed for all other DR systems (results are not shown), although it was less pronounced for the Sectra system. The DQE of this system remains quite constant with dose, due to the low structural noise [30] and the electronic threshold technique associated with the photon counting system [27]. The maximum values of the DQE reached and the
corresponding dose are summarised in Table 7.5, for all the DR systems. The results of the photon counting system are also shown.

At a relative very low dose, the Sectra system has the highest DQE value of all the DR systems examined. The DQE values of the GE and Hologic systems are comparable, although for the direct DR system this was achieved with a significant reduction in dose. With a relative high dose, the two other direct DR systems have the lowest DQE of all the systems. Considering the uncertainty of the DQE calculations the difference in the DQE values of the DR systems is, however, quite small. The DQEs of the CR systems were computed at the diagonal direction (preMTF(u) and preMTF(v) were averaged in this case) and plotted as a function of spatial frequency for the corresponding reference dose, as shown in Fig.7.15.

Table 7.5. Maximum value of the DQE for each system investigated. The corresponding dose and spatial frequency are also shown

<table>
<thead>
<tr>
<th>System</th>
<th>Max DQE (%)</th>
<th>Dose (μGy)</th>
<th>Frequency (mm⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hologic Selenia</td>
<td>59</td>
<td>114</td>
<td>0.67</td>
</tr>
<tr>
<td>GE Essential</td>
<td>64</td>
<td>135</td>
<td>0.17</td>
</tr>
<tr>
<td>Giotto IMS</td>
<td>54</td>
<td>171</td>
<td>0.19</td>
</tr>
<tr>
<td>Siemens Inspiration</td>
<td>56</td>
<td>275</td>
<td>0.73</td>
</tr>
<tr>
<td>Sectra MDM-L30</td>
<td>69</td>
<td>46</td>
<td>0.32</td>
</tr>
</tbody>
</table>

The needle plate CR (Konica CP-1M) has the highest DQE of the all other CR systems. This is due to the high resolution associated with the needle crystal technology by which the lateral light signal scatter is extremely minimised. The two other Konica plates have the second highest maximum DQE (53 %) after the needle Konica CR plate, but the relative poor resolution of the plate RP 6-M makes the DQE drop at middle to high frequency. The DQE of the Agfa CR system is as good as that for the Konica RP-7M at the middle frequency. This is due to the relative low noise associated with this system. Its DQE, however, falls off at high frequency as this system has the lowest MTF of all other systems. The sharp drop in the low-frequency DQE of the Agfa and Kodak systems is presumably related to the significant variation in their NNPS between very low- and very high-frequency, due to uncorrected non-uniformities. The other reason is that their MTF was measured using PMMA added filtration which usually generates a relative large LFD. The Fuji CR has the lowest DQE curve of the all other CR systems, at the middle and high frequency range. The
relative highest reference dose of this system, of all other system, may has an effect on its DQE which decreases with dose for the CR technology.

The dependence of the DQE of the CR systems on exposure was investigated using the Fuji CR system. The DQE of this system was computed in the diagonal direction and plotted as a function of spatial frequency for a range of dose levels, as shown in Fig. 7.16. The DQE of the system decreases with an increase in X-ray dose, due to the presence of the significant amount of structural noise which is proportional to the square of exposure. Fig. 7.17 shows the maximum DQE values plotted as a function of the entrance air kerma for the GE, Siemens and Fuji CR systems. For the CR system the DQE is continuously decreasing as dose increases, over the whole range of dose. This is due to the uncorrected large amount of structural noise. For the GE and Siemens DR systems, in contrast, this behaviour occurs only from about 300 μGy and 400 μGy, respectively. This is expected for the DR system in which the structural noise is reduced using a flat field procedure. Therefore, the DQE of these systems increases with dose up to a certain level of dose at which the structural noise (proportional to dose) becomes more important than the quantum noise (proportional to the square root of dose). The performance of the DQE of the investigated CR and DR systems was observed (results are not shown) for all the other systems of the same technique of detection. A similar DQE behaviour was also reported in a previous study [27], for CR and DR systems.

7.3.3 Comparing the performance of systems

The threshold gold thicknesses measured at different detail diameters on the GE Essential system are plotted in Fig. 7.18. The fitted curves were used to determine the doses required to reach the acceptable and achievable standards (in terms of threshold gold thicknesses), as shown in Fig. 7.19 for the 0.1 mm detail diameter. Figs. 7.20 and 7.21 show the doses required to reach the acceptable and achievable gold thickness thresholds of all detail diameters for the GE DR and Kodak CR systems, respectively. A similar procedure was followed for each system (results are not shown).
Fig. 7.15 DQE(\(\mu, v\)) of the CR systems calculated at the reference dose level of each system. The error bars indicate the 2 sem of the measurement.

Fig. 7.16 DQE(\(\mu, v\)) of the Fuji Capsula CR system measured for a range of levels of dose. The error bars indicate the 2 sem of the measurement.
The dose required to meet the image quality standards for the GE system was about 0.5 mGy for the acceptable level and just over 1.0 mGy at the achievable level. These dose requirements were similar at all detail sizes considered. For the Selenia system the dose required to meet the acceptable image quality standard was about 0.6 mGy, but about 1.5 to 2.5 mGy for the achievable level. This system seemed to need slightly lower dose levels for the smaller details than those required for the larger details. This may be explained by the relatively good MTF for the Selenium detector used by this system. The dose required to meet the image quality standards for CR systems was very dependent on the detail size considered. Thus for the Kodak CR system a dose of 2.38 mGy was required at the acceptable image quality level at the 0.1 mm detail size. At the 0.5mm detail size the dose at the acceptable level would be 1.41 mGy. It is also noted that for most CR systems the achievable image quality cannot be reached for doses less than the upper acceptable limit in the guidelines of 3 mGy. This is the limit that applies for breast or phantoms equivalent to a 50mm thickness of PMMA [8].

The fact that dose requirements are higher for CR systems for small details is presumably due to their relatively low MTF at higher spatial frequencies. The Konica system was tested with three different plates with different MTFs. The plate with the
best MTF and DQE (CP-1M), also required the lowest dose for detecting the smallest details.

![Curves fitted]

\[ T = D^{-n}; n = 0.51 \pm 0.02 \]

**Fig. 7.18** Threshold gold thickness measured on the GE Essential system at 5 dose levels

**Fig. 7.19** Determination of the doses required to reach the minimum and achievable standards for 0.1 mm detail diameter for the GE system. The red solid and the blue dash lines indicate the dose limit required to achieve the acceptable and the achievable image quality, respectively. The error bars indicate the 2 sem of the measurement.
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Fig. 7.20 The MGD calculated to be necessary to reach the achievable and acceptable image quality levels in European Guidelines at different detail sizes for the GE Essential system (Determined using predicted human reading of the CDMAM images.)

Fig. 7.21 Dose required to meet the acceptable and achievable image quality levels in European Guidelines for the Kodak CR system. (Determined using human reading of the CDMAM images.)
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Figs. 7.22 and 7.23 show the dose required to meet the acceptable image quality limits in the European Guidelines and the DQE at corresponding detail sizes and spatial frequencies. It can be observed that the dose required to meet the acceptable requirements in European Guidelines was, as might be expected, strongly related to the DQE of the detector systems. Thus where a system has a relatively low DQE at high spatial frequencies (e.g. 5 mm$^{-1}$) a relatively large dose is required to meet the minimum image quality level in European Guidelines for the smallest detail size (i.e. 0.1 mm). The curves in these plots are divided into two regions in terms of DQE and the dose required to meet the image quality limits. In general, the CR systems are found in the upper left region whereas the DR systems are located at the lower right region. However, the use of the needle CR plate will move the CR system as close as possible to the DR region, in which a relatively small dose is required to meet the minimum image quality standards specified in the European Guidelines.

![Graph showing MGD (mGy) vs. DQE at 0.1 mm detail size and 5 mm$^{-1}$ spatial frequency.](image)

Fig. 7.22 Plot of MGD (mGy) required to meet the minimum image quality (IQ) levels in the European Guidelines and the DQE at 0.1 mm detail size and 5 mm$^{-1}$ spatial frequency. The fitted curves assume an inverse relationship
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Fig. 7.23 Plot of MGD (μGy) required to meet the minimum image quality (IQ) levels in the European Guidelines and the DQE at 0.25 mm detail size and 2 mm⁻¹ spatial frequency. The fitted curves assume an inverse relationship.

7.3.4 Contrast detail measurements and the theory model

The modelled threshold contrast was obtained for all systems using Eq. 7.2. In order to determine the DQE at zero spatial frequency, the data points were first smoothed using a moving average filter with a window of 5. A first-order polynomial extrapolation was then applied to the data points located between 1.7 and 0.9 mm⁻¹. This is because the measured data points of DQE are often affected by low frequency artefacts present in the NPS image [31]. Table 7.6 gives the extrapolated DQE (0) and the calculated noise equivalent apertures for each system investigated. Figs. 7.24 - 7.29 show the experimental and theoretically derived contrast detail curves for the Konica CR (CP-1M, RP-6M and RP-7M), Fuji CR, GE Essential and Siemens Inspiration systems, respectively. For all curves the experimental data indicates the fit to predicted threshold gold thickness at the corresponding dose levels using automated reading of the CDMAM images (Eq. 5.3 and 5.6).

For all the systems investigated, the modelled contrast detail results have a very good fit to those obtained experimentally. The measured results are well predicted by the model with an average variation of approximately ± 8%. For the large details (0.4 -
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1.0 mm), however, the model gives threshold gold thicknesses which are lower than the measured data by approximately 18%, on average. This is presumably, due to the presence of the inter- and intra-observer errors which are not included in the model. For the GE and Fuji CR systems there is a noticeable increase in the modelled threshold gold thickness at the smallest disk diameters (0.08 mm and 0.1 mm). This is likely to be due to the low MTF of these two systems at the high spatial frequencies. It is important to note that the threshold contrast is influenced by the visual processing mechanisms and the inherent unsharpness of the imaging device at large and small detail diameters, respectively [5].

Table 7.6. The value of DQE at zero spatial frequency and the NEA for each system investigated

<table>
<thead>
<tr>
<th>System</th>
<th>NEA (mm²)</th>
<th>DQE (0) (%)</th>
<th>Half dose</th>
<th>Normal dose</th>
<th>Double dose</th>
</tr>
</thead>
<tbody>
<tr>
<td>Siemens Inspiration</td>
<td>0.045</td>
<td>38</td>
<td>48</td>
<td>55</td>
<td></td>
</tr>
<tr>
<td>GE Essential</td>
<td>0.085</td>
<td>57</td>
<td>53</td>
<td>53</td>
<td></td>
</tr>
<tr>
<td>Fuji CR</td>
<td>0.097</td>
<td>38</td>
<td>37</td>
<td>33</td>
<td></td>
</tr>
<tr>
<td>Konica CR CP-1M</td>
<td>0.074</td>
<td>69</td>
<td>65</td>
<td>58</td>
<td></td>
</tr>
<tr>
<td>Konica CR RP-6M</td>
<td>0.098</td>
<td>56</td>
<td>51</td>
<td>40</td>
<td></td>
</tr>
<tr>
<td>Konica CR RP-7M</td>
<td>0.085</td>
<td>57</td>
<td>51</td>
<td>42</td>
<td></td>
</tr>
</tbody>
</table>

Considering the simplicity of the model used there is a very good fit to the experimental data. However, further modification of the model appears necessary in order to take account of observer visual response, geometric blurring and heel effect.

Fig. 7.24 Experimental (points) and theoretical (lines) contrast detail measurements obtained at three levels of dose for the CP-1M plate of the Konica CR system. Error bars indicate 95% confidence limits.
Fig. 7.25 Experimental (points) and theoretical (lines) contrast detail measurements obtained at three levels of dose for the RP-6M plate of the Konica CR system. Error bars indicate 95% confidence limits.

Fig. 7.26 Experimental (points) and theoretical (lines) contrast detail measurements obtained at three levels of dose for the RP-7M plate of the Konica CR system. Error bars indicate 95% confidence limits.
Fig. 7.27 Experimental (points) and theoretical (lines) contrast detail measurements obtained at three levels of dose for the Fuji CR system. Error bars indicate 95% confidence limits.

Fig. 7.28 Experimental (points) and theoretical (lines) contrast detail measurements obtained at three levels of dose for the GE Essential system. Error bars indicate 95% confidence limits.
Fig. 7.29 Experimental (points) and theoretical (lines) contrast detail measurements obtained at three levels of dose for the Siemens Inspiration system. Error bars indicate 95% confidence limits.

7.4 Conclusions

In this chapter both objective and subjective measurements were used to evaluate and compare image quality of a wide range of CR and DR digital mammography systems. The contrast detail detectability of these systems was determined and analysed across a range of doses and the results were presented as the dose required to achieve the standards in the European guidelines at different detail sizes. MTF, NNPS, NEQ and DQE were also obtained for the detectors used in these systems following the IEC protocol.

It was found that the dose required to meet the image quality standards for the indirect DR system was about 0.5 mGy for the acceptable level and just over 1.0 mGy at the achievable level. These dose requirements were similar at all detail sizes considered. For the direct DR Selenia, on the other hand, the dose required to meet the acceptable image quality standard was about 0.6 mGy, but about 1.5 to 2.5 mGy for the achievable level. This system seemed to need slightly lower dose levels for the smaller details than those required for the larger details. This may be explained by the relatively good MTF for the selenium detector used by this system. The dose required
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to meet the image quality standards for the CR systems was found to be very
dependent on the detail size considered (2.38 mGy and 1.41 mGy were required at the
acceptable image quality level at the 0.1 mm and 0.5 mm detail sizes, respectively). It
was also observed that for most CR systems the achievable image quality cannot be
reached for doses less than the upper acceptable limit in the guidelines of 3 mGy. The
fact that dose requirements are higher for CR systems for small details is presumably
due to their relatively low MTF at higher spatial frequencies, as the threshold contrast
of small detail diameter is usually related to the MTF of the imaging device. The
Konica CR system was tested with three different plates with different MTFs. The
plate with the best MTF and DQE (CP-1M), also required the lowest dose for
detecting the smallest details.

Though the indirect DR system has the lowest high-frequency NNPS, its NNPS curve
was not constant over approximately all frequencies, possibly due to the low MTF of
this version of the GE system. For direct DR systems (including the photon counter)
the NNPS curves were relatively uniform over a large spatial frequency band. This is
presumably a result of their MTF which remains high up to the Nyquist frequency.
For the all DR systems both MTF and NNPS were reasonably isotropically radial.
While the NNPS of the photon counting system was symmetric in both directions, the
MTF was significantly lower in the slit scan direction than that in the array direction.
The CR systems showed, as expected, significant asymmetric MTF and NNPS
between the scan and subscan directions.

At the clinical exposure levels the DQE of the DR systems was found to be dose
independent. For the CR system, in contrast, the DQE decreased with an increase in
X-ray dose. This is due to the presence of the huge amount of structural noise, which
is proportional to the exposure. However, the use of the crystal needle CR plate would
improve the DQE of the CR system, at the clinical doses.

This chapter has considered whether the performance determined by contrast detail
measurement is correlated with that anticipated from the results on detector
performance. As expected, there was a strong relationship between the dose required
to meet the acceptable requirements in European Guidelines and the DQE of the
detector systems. Thus a relatively large dose required to meet the image quality
limits is expected for a system with a relatively low DQE. In practice for a given
system this relationship will depend on some other factors such as the scatter rejection method and beam quality.

Finally, the feasibility of using the objective measurements to predict the contrast detail measurements was further investigated using a wide range of CR and DR digital mammography units. This was done using a simple signal-matched noise-integration model. Considering the simplicity of the model used there was a very good fit to the experimental data. A more refined model appears necessary in order to take account of several ambient viewing conditions. These include geometric blurring, heel effect, scattered radiation and observer visual system. However, a successful model has the potential to predict contrast detail performance from standard measurements of detector performance and may provide a more reproducible method of comparing the performance of digital mammography systems. A more sophisticated version of this model could provide an alternative method of evaluating the performance of digital mammography systems against European standards.

References


7. Stefano Rivetti, Nico Lanconelli, Renato Campanini, Marco Bertolini, Gianni Borasi, and Andrea, Claudio Danielli, Lidia Angeli and Stefania Maggi, Comparison of different commercial FFDM units by means of physical characterization and contrast-detail analysis, Medical Physics 33 (11), 2006, pages: 4198-4209.

CHAPTER 7. PREDICTING CONTRAST DETAIL DETECTIBILITY


25. J. A. Segui and Wei Zhao, Amorphous selenium flat panel detectors for digital mammography: Validation of a NPWE model observer with CDMAM observer performance experiments, Medical Physics 33 (10), 2006, pages: 3711 – 3722
CHAPTER 7. PREDICTING CONTRAST DETAIL DETECTIBILITY


Chapter 8

Simulation of Image Performance of CZT Detector for Digital Mammography*

8.1 Introduction

Direct conversion solid-state detectors that convert the incident X-ray photons directly into electron-hole pairs are now under development [1,2] for application to digital mammography. These technologies are operated using active matrix imagers such as silicon (Si), selenium (Se), cadmium telluride (CdTe), cadmium zinc telluride (CZT), gallium arsenide (GaAs) and lead iodide (PbI₂). These materials are ideal for mammography detectors, because they have high X-ray absorption efficiency (except Si) and extremely high intrinsic spatial resolution and low noise [3].

Because of its availability in relative large sizes and high bulk resistivity, the CZT detector has promise for medical imaging purposes. Moreover, CZT has a high atomic number and high density, which make it possible to construct a very thin X-ray detector that still has high quantum interaction efficiency.

In chapters 6 and 7 the concepts of MTF, NPS and DQE were discussed and studied experimentally for the evaluation of image performance of CR, α-Si, α-Se and photon counting multi-slit scanning systems. In this chapter, a CZT detector is evaluated, with the aid of the Monte Carlo (MC) technique, for digital mammographic application.

In order to evaluate the proposed detector, a simulation of X-ray photon transport in the sensor material was conducted using the extended version of Monte Carlo N-

CHAPTER 8. IMAGE PERFORMANCE OF CZT DETECTOR

Particle (MCNPX) code [4]. The code tracks both photons and electrons with an energy cut-off of 1keV after which the particle is killed. Photoelectric absorption, coherent and incoherent scattering are included in the physics of photon transport of the code. It also accounts for X-ray fluorescence effects. Charge carrier transport was simulated through the solution of Poisson's equation and the assumption of fully depleted semiconductor detectors. The two models were then combined in order to evaluate the response of the detector to the line source.

The presampled MTF, NPS and DQE are then obtained for a pixelated CZT detector. Results of a CdTe detector are also presented to verify the difference (if any) between the two detectors. GaAs, Si and PbI₂ detectors are also included in the study for comparison.

8.2 Materials and Methods

8.2.1 Detector Characterisations

One important property of a material in radiation detection can be investigated by noting its energy resolution which is defined as the response of a detector to a monoenergetic radiation source [5]. The energy resolution of the CZT detector was simulated using the pulse height tally (Tally "F8) and the Gaussian energy broadening (GEB) function provided by the MCNPX code. The GEB function is used for better simulation of a physical radiation detector in which energy peaks follow Gaussian distribution [4]. Using this option, FWHM of the broadened energy (\(E\)) can be specified as,

\[
FWHM = a + b\sqrt{E + cE^2}
\]  

[8.1]

The units of \(a\), \(b\), and \(c\) are MeV, MeV\(^{1/2}\) and none, respectively.

To validate the simulation work, a 3 mm × 3 mm × 1 mm single CZT crystal was first characterised experimentally using \(^{55}\)Fe and \(^{241}\)Am radiation sources. Fig.8.1 shows the detector geometry used in the experimental work. The detector is placed into a vacuum covered by 1 mm aluminium and 0.1 mm beryllium window. The crystal-to-window distance is about 2 mm. The detector was then simulated using this geometry and irradiated with sources of the same energy peaks of \(^{55}\)Fe and \(^{241}\)Am. From the
experimental work, three different FWHMs were obtained in order to solve Eq.8.1. The simulated energy spectra of the two sources was then obtained and validated against the experimental work.

Fig.8.1. The detector geometry used in the simulation work. The subdivided levels, used in the MTF work, are also shown

**8.2.2 Mammography Energy Spectrum**

Fig.8.2 shows the simulated energy spectra used in this work. The spectrum was first generated using the programme provided by the IPEM [6], in the typical mammography conditions. These are a Mo/Mo target/filter combination with an anode angle of about 25° and a 0.127 μm thick beryllium window and a maximum energy of 28 kVp. The spectrum was regenerated by the MCNPX code in order to verify the source spectrum definition and the two spectra are in good agreement as shown in Fig.8.2. The latter one was used for the all following measurements.

**8.2.3 Detector Response**

To assess a medical imaging system in terms of MTF and NPS, the system has to be linear and shift invariant [7]. It is well known [8] that digital detectors have a linear response to the input radiant exposure. On the other hand, they are shift-variant systems which means that their MTF and NPS depend on the position of the input radiation source, i.e. phase dependent. The difficulties in the MTF measurements for the shift-variant detectors and the solutions to these difficulties were discussed in
CHAPTER 8. IMAGE PERFORMANCE OF CZT DETECTOR

chapters 2 and 6. For the NPS analysis all systems used here were assumed to be shift-invariant as discussed earlier.

In order to measure the detector response, a collimated beam with the X-ray mammography spectrum located at 65 cm from the detector, was simulated and used to obtain uniform images, 128 pixel x 128 pixels each. The beam was transmitted through a slab of modelled breast with 53 mm thickness (the average thickness of a compressed breast in the clinical application) placed halfway between beam source and the detector. The breast tissue composition used here was provided by the International Commission on Radiation Units and Measurements (ICRU) report 44 [9]. Fig.8.3 shows the original and the transmitted X-ray spectra. The latter was used as the detector incident spectrum to obtain the average number of photons (q) per unit of area striking the detector. This was done using Tally F2 provided by the MCNPX code. The average number of the photons incident on the detector was then converted into air kerma using tabulated data [6]. The photon flux per unit entrance air kerma was found to be 5348 photons/μGy/mm² for a spectrum with a beam quality of 0.6012 mm aluminium. The mean pixel value was measured in a 5 mm x 5 mm ROI located in the centre of the uniform image as shown in Fig.8.4. This was repeated for a range of entrance air kerma; 10, 40, 82, 163 and 326 μGy. The detector response was then obtained by plotting the mean pixel values against the doses.
8.2.4 Transport Models

Using the Monte Carlo (MC) technique, it is easy to vary the detector size and its geometric set up. The MC technique was used here to model a hybrid pixel detector for digital mammography imaging. However, to simulate the response of the semiconductor detector to an incident radiation stimulus, both the photon and charge carrier transports need to be modelled.

Fig. 8.3 shows the original (source) and the attenuated (input) Mo/Mo X-ray spectra. The latter was used for the detector response, NPS and DQE measurements.

I. Photon Transport Model

The X-ray photon transport in semiconductor sensors has been modelled in the literature [10-13] using different MC techniques. Here, the MCNPX code was used to simulate the photon transport in the CZT detector material. Appendix B shows the full MCNPX file created and used in this work.
Fig. 8.4 shows the ROI in the centre of a uniform image obtained at 10 μGy incident air kerma, using the CZT detector. This image is also an example of the simulated images used for NPS analysis.

To model the X-ray interactions, a 3 mm × 3 mm × 0.5 mm CZT detector was simulated. This was divided up into levels with 100 μm in depth, segmented into 100 μm square pixel size and irradiated with a line source of the mammography energy spectrum shown in Fig. 8.2. The source was 20 μm in width, slightly tilted with respect to the pixel array and impinging on the centre of the detector. For all other detectors used here the same configuration and geometry of the simulated CZT detector were used.

The deposited energy distribution over the pixellated detector was tallied (counted), using Tally "F8. The number of charge carriers created in each pixel was then calculated as,

\[ \frac{E_d}{E_g} \]  

(8.2)

where \( E_d \) is the deposited energy and \( E_g \) is the minimum energy required to generate one electron-hole pair within a particular detector.
II. Charge Transport Model

There are several well-established and commercially available simulation tools for charge transport in semiconductor devices. The most common one is the tool using the so-called drift-diffusion charge transport model [11]. In this model, the solution of the continuity, the drift-diffusion transport and the Poisson equations provide averaged quantities of the carrier concentration and velocity and electric field distribution, etc. However, for simple semiconductor materials, the analytical method can be used by applying Poisson's equation and the depletion approximation [10]. In this work, the analytical approach was adopted as described below.

When a $p$-type (or $n$-type) crystal is doped with a uniform concentration of acceptor (or donor) impurity, the free electrons and holes start to diffuse into regions with lower concentrations of electrons ($p$-side) and holes ($n$-side). Following the diffusion, the electrons and holes come into contact with each other and vanish by a process known as recombination. This process will end up with fixed negative charges on the $p$-side and fixed positive charges on the $n$-side of the $p-n$ junction. The region over which the fixed charges present is called the space-charge region or the depletion region as shown in Fig.8.5.

Consequently, an electric potential difference builds up across the depletion region in the direction that it just resists the further diffusion of electrons and holes across the junction. The value of the potential $\varphi$ at any point can be found by solution of Poisson's equation [5],

$$\nabla^2 \varphi = -\frac{\rho}{\varepsilon} \quad (8.3)$$

where $\rho$ is the net charge density and $\varepsilon$ is the dielectric constant of the semiconductor. Where a potential gradient exists, there must also be an electric field $E \ (V/m)$ in the opposite direction to that potential gradient or simply as,

$$E = -\varphi \quad (8.4)$$
Substituting Eq.8.3 by Eq.8.4, the electric field in x-dimension \( E(x) \) can be obtained by,

\[
E(x) = \frac{\rho(x)}{\varepsilon}
\]  

(8.5)

When a sufficient external reverse bias voltage is applied to the \( p-n \) junction of the detector, the space-charge region at the junction becomes effectively wider and a \textit{fully depletion} region is created. The charge density in the depth axis \( \rho(x) \) over the depleted region on the \( n \) side of the junction is given by,

\[
\rho(x) = eN_D
\]  

(8.6)

where \( e \) is the electron charge and \( N_D \) is the donor concentration in the semiconductor.

Substituting Eq.8.5 by Eq.8.6 gives,

\[
E(x) = \frac{eN_D}{\varepsilon} (x - d)
\]  

(8.7)

where \( d \) is the entire distance over which the space charge extends (\( \approx \) the detector thickness in the fully depleted semiconductor).
CHAPTER 8. IMAGE PERFORMANCE OF CZT DETECTOR

When X-ray interactions take place within the depletion region of a semiconductor detector, the created charge carriers (electron-hole pairs) will be drifted out of the depletion region by the applied electric field and diffuse from one side. At low-to-moderate values of the electric field intensity (before the saturation point is reached), the drift velocity \( V \) of the carriers is proportional to the electric field in the fully depleted semiconductor, so;

\[
V_h = \mu_h E \\
V_e = \mu_e E
\]

(8.8)

(8.9)

where \( \mu_h \) and \( \mu_e \) are hole's and electron's mobility, respectively. The drift velocity of a hole in the depth coordinate is given by substituting Eq.8.7 by Eq.8.8 as,

\[
V(x) = \frac{\mu_h eN_D}{\varepsilon} (x - d)
\]

(8.10)

The corresponding drift time is obtained by integrating the inverse of the drift velocities \( V \) from the origin of the charge generation as,

\[
t(x) = \int_{x_0}^{x} \frac{1}{V(x)} dx
\]

(8.11)

Substituting Eq.8.10 by Eq.8.11 gives,

\[
t(x) = -\frac{\varepsilon}{\mu_heN_D} \ln \left| -\frac{x}{(1 + a/2)d} \right|
\]

(8.12)

where \( a \) is the so-called depletion coefficient which is defined as the ratio of the bias voltage to the built-in voltage (the voltage created in the depletion region of the junction) and it equals 0 in the assumption of fully depleted semiconductor [10]. Table 8.1 shows the physical properties of the semiconductor materials used in this work.

With their drift only, all charge carriers would travel along a line, corresponding to the applied field, from the generation point to the collection point. However, this is
CHAPTER 8. IMAGE PERFORMANCE OF CZT DETECTOR

not the case, the charge carriers are influenced by lateral diffusion due to random thermal movement. At a given time, this diffusion introduces some spread in the collection position which can be characterised as a Gaussian distribution whose FWHM is given by,

\[ FWHM = 2.355\sigma \]  

(8.13)

where \( \sigma \) is the standard deviation which is given [5] by,

\[ \sigma = \sqrt{2Dt} \]  

(8.14)

where \( D \) is the diffusion coefficient and \( t \) is the elapsed time. The value of \( D \) can be predicted from;

\[ D = \mu \frac{kT}{e} \]  

(8.15)

where \( T \) is the absolute temperature and \( k \) is the Boltzmann constant. At room temperature, the value of \( kT/e \) equals 0.0253 \( V \). Fig.8.6 shows the values of FWHM plotted as a function of depth for the semiconductor detectors used here.

Table 8.1. Physical properties of the semiconductor detectors used in this work

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Si</th>
<th>CZT</th>
<th>CdTe</th>
<th>GaAs</th>
<th>PbI₂</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average atomic number</td>
<td>14</td>
<td>49.1</td>
<td>50</td>
<td>31.5</td>
<td>63</td>
</tr>
<tr>
<td>Density (g cm⁻³)</td>
<td>2.33</td>
<td>5.78</td>
<td>5.85</td>
<td>5.32</td>
<td>6.2</td>
</tr>
<tr>
<td>Electron mobility (cm²V⁻¹s⁻⁻)</td>
<td>1400</td>
<td>1000</td>
<td>1100</td>
<td>8000</td>
<td>8</td>
</tr>
<tr>
<td>Hole mobility (cm²V⁻¹s⁻⁻)</td>
<td>450</td>
<td>120</td>
<td>100</td>
<td>400</td>
<td>2</td>
</tr>
<tr>
<td>Electron diffusion coefficient (cm²s⁻⁻)</td>
<td>37.55</td>
<td>29.10</td>
<td>29.78</td>
<td>201.20</td>
<td>0.202</td>
</tr>
<tr>
<td>Hole diffusion coefficient (cm²s⁻⁻)</td>
<td>11.63</td>
<td>1.64</td>
<td>2.80</td>
<td>10.36</td>
<td>0.051</td>
</tr>
<tr>
<td>Energy bandgap (eV)</td>
<td>1.12</td>
<td>1.57</td>
<td>1.44</td>
<td>1.43</td>
<td>2.32</td>
</tr>
<tr>
<td>Pair creation energy (eV)</td>
<td>3.62</td>
<td>4.64</td>
<td>4.43</td>
<td>4.20</td>
<td>4.9</td>
</tr>
<tr>
<td>Relative dielectric constant*</td>
<td>11.70</td>
<td>10.00</td>
<td>10.90</td>
<td>12.80</td>
<td>21.0</td>
</tr>
<tr>
<td>Fano factor</td>
<td>0.10</td>
<td>0.082</td>
<td>0.20</td>
<td>0.18</td>
<td>0.14</td>
</tr>
<tr>
<td>Resistivity (Ω/cm)</td>
<td>&lt;10⁴</td>
<td>3×10⁹</td>
<td>10⁹</td>
<td>10⁹</td>
<td>10¹³</td>
</tr>
</tbody>
</table>

* Relative to absolute dielectric constant of vacuum which equals 8.854x10⁻¹² F
The broadening of the charge carrier distribution in the arrival position would give a limitation to the position measurement in semiconductor detectors. This limitation was taken into account by calculating the value of $\sigma$ for the drift times corresponding to the depth point at which a charge cluster is generated. This means the $x$ and $y$ position of the charge cluster at the collector's plane, obtained by the MCNPX simulation, was randomly blurred according to the value of $\sigma$. This was done using a code written in the Matlab programme [14], and shown in Appendix C. In addition, the generation of the charge carriers is also subjected to the statistical fluctuation or so-called Fano noise. This means that the observed variance in the number of charge carriers can be predicted [5] by,

$$F = \frac{\text{observed variance in } N}{\text{Poisson predicted variance } (= N)} \quad (8.16)$$

where $N$ is the number of charge carriers generated in each individual X-ray interaction and $F$ is the Fano factor, which has been measured for the semiconductors

![Fig. 8.6. Values of FWHM plotted against thickness of CZT, CdTe, Si, GaAs and PbI$_2$ semiconductor detectors.](image-url)
CHAPTER 8. IMAGE PERFORMANCE OF CZT DETECTOR

traditionally employed for application of X-ray detection [15,16] and presented in Table 8.1.

8.2.5 MTF

Digital detectors are often composed of discrete pixels. The dimension of the active area of each detector pixel defines the aperture of that detector. The aperture response of the detector is then one inside the aperture and zero outside. Fig. 8.7 shows the aperture response for two different active pixel sizes; 50 and 100 µm and their corresponding MTF\textsubscript{aperture} calculated by Eq.2.8.

Fig.8.7 shows the pixel aperture response for pixel size of 50 µm (above-left) and 100 µm (above-right), and their corresponding 2D-MTFs (below)

Fig.8.8 shows the simulated 2-D image of the slightly tilted line source obtained using the pixelated CZT detector. The line spread function (LSF) was estimated by
acquiring 90° signal profiles across the image. The row data of the image was corrected by normalizing the values of each row to the sum of that row to compensate the variations (if any) along the line source. A series of individual signal profiles along the rows were combined together (with ± 9.4% standard deviation) to obtain the sampled LSF. The adequate number of these individual LSFs, required for synthesising the sampled LSF, was determined by plotting the pixel values along the vertical direction (columns in the row data). 19 individual LSFs were found to be sufficient for estimating the sampled LSF, as shown in Fig. 8.9. The sampled LSF was computed by plotting the value in each pixel versus the distance from the centre of the line source (L in fig. 8.8).

MTF was computed by taking the modulus of the 1D fast Fourier transform (FFT) of the sampled LSF and normalizing its value to 1 at zero frequency. This is the sensor MTF which was then multiplied by the aperture MTF to give the preMTF of the detector.

Fig. 8.8. The simulated image of a 20 μm line source on top of the CZT detector. Note that for the display purpose only, the line source was tilted with 15° with respect to the detector array.

8.2.6. NPS

Five 128 pixels × 128 pixels uniform images were simulated and used to compute the NPS analysis. The images were simulated using the same method and conditions used earlier for the detector response measurement. Fig. 8.4 shows an example of these
images used for the NPS calculation. The images were then Fourier transformed, squared and averaged to give the 2-D NPS of the detector. This was then multiplied by the squared pixel size and divided by the number of pixels, in the two dimensions, to express the NPS in units of area (mm²), as per Eq.2.9. Finally, the NPS value was divided by the squared mean pixel value of the uniform image to give the normalised NPS (NNPS). This was repeated for each dose mentioned in section 8.2.3.

**Fig. 8.9.** The pixel values along the vertical direction used to provide the adequate number of rows of the row data of the LSF image needed to obtain the sampled LSF

In fact, NPS is usually calculated from uniform images after subtraction of the non-stochastic background signal (added noise) [17]. However, the noise in the simulated uniform images used here represents only the quantum noise and therefore there was no need to perform the background subtraction.

To calculate DQE, 1D NPS is required. This was obtained by averaging the vertical lines in the 2D NPS (4096 digital values with ±5.8 % standard deviation) which are then grouped into frequency bins of 0.078 mm⁻¹ and plotted up to the Nyquist frequency (5 lp/mm).

### 8.2.7. DQE

The DQE was calculated as,

\[
DQE = \frac{MTF^2}{NNPS \cdot q(K_a)}
\]  

(8.17)
where MTF is the presampled MTF, NNPS is the normalised noise power spectrum and \( q \) is the average number of photons incident on the detector surface per unit area and unit exposure, in air kerma \( (K_a) \). \( q \) was obtained for each level of dose as explained in section 8.2.3. The DQE was then computed for all detectors used here.

8.3 Results

8.3.1 Detector Characterisations

The total linear absorption coefficient for the CZT detector and all its main components was calculated as a function of incident photon energy, using the MCNPX code and presented in Fig.8.10. The quantum detection efficiency, which is defined as the percentage of absorbed X-ray photons in the detector material to the total incident flux, was obtained for the CZT detector and plotted with comparison with Si and CsI (TI) detectors in Fig.8.11. For better comparison the Si detector was simulated with thickness similar to that of the CZT detector. The thickness of the phosphor (CsI (TI)) employed in the commercially available \( a-Si \) digital mammography, was used in this comparison. Fig.8.12 shows the experimental and simulated energy spectra of the \( ^{55}Fe \) and \( ^{241}Am \) for the CZT detector.

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![Graph showing components and total cross-section of CZT detector](image)

**Fig.8.10 Components and the total cross-section of CZT detector plotted against the photon energy range 1 keV – 10^7 MeV (produced by the MCNPX code)**

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Fig. 8.11. Absorption energy calculated as a function of incident energy of the CZT, Si and CsI (TI) detectors using the MCNPX code. The thickness of the semiconductor detectors was chosen to be the same for better comparison. For scintillator detector the clinical thickness was used.

Fig. 8.12. Measured and simulated spectra of gamma ray from $^{55}$Fe (5.9 keV) and X-rays from Np$_{cal}$ (13.9 keV) and Np$_{pl}$ (59.5 keV) of the CZT detector. Note that the MCNPX code produces pulses normalised to the source particle which gives a spectrum with high counts as large number of particles (histories) are required to give results within the accepted errors (<10%).
8.3.2 Detector Response

The linearity of the CZT detector was verified and is shown in Fig. 8.13. Mean pixel value was measured for each uniform image and plotted as a function of entrance air kerma at which the image was acquired.

![Graph showing linearity of CZT detector](attachment:graph_image.png)

Fig. 8.13 shows the linearity of the CZT detector plotted as the average pixel values versus entrance air kerma.

8.3.3 MTF

Fig. 8.14 shows the sampled LSF of the CZT detector normalised to the maximum value of one. The sensor MTF of the CZT detector calculated by taking the modulus of the 1D FFT of the sampled LSF and normalizing its value to 1 at zero frequency, is shown in Fig. 8.15.

Because the preMTF of the digital detector is mainly determined by the aperture size of the detector, it is better to make the comparison between the sensor materials using the sensorMTF instead of the preMTF. The sensorMTFs of CdTe, Si, GaAs and PbI₂ were modelled and are shown in Fig. 8.17 as a comparison with the CZT detector. The horizontal line in Fig. 8.17 indicates the 10% level of the MTF curve which corresponds to the resolving power of the digital system [10].

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Fig. 8.14. The sampled LSF of the CZT detector. Note that the relative error associated with the simulation work is ±10.2% and the standard deviation of the LSFs averaging is ±9.4%.

Fig. 8.15. The sensor MTF measured as a function of spatial frequency in the plane of the charge collection of the CZT detector. The error bars represent the relative error of the simulation work.

Fig. 8.16 shows the preMTF of the 100 µm pixel size CZT detector obtained as the product of the sensor MTF and the aperture MTF.

In fact, the performance of the digital semiconductor detectors depends on several factors. These include carrier mobility and life time, material structure, trapping impurities, temperature, dopant concentration [18] and sensor thickness. The effect of
the last two factors on the sensor MTF was investigated. Fig. 8.18 shows the sensor MTF obtained for the CZT detector with various thicknesses. Fig. 8.19 shows the sensor MTF obtained at different donor concentrations for the GaAs detector.

![Graph showing sensor MTF](image)

**Fig. 8.16.** The preMTF of the 100 μm pixel size CZT detector measured up to the Nyquist frequency.

![Graph comparing sensor MTFs](image)

**Fig. 8.17.** The sensor MTFs of the CZT, CdTe, Si, GaAs and PbI₂ detectors. The horizontal line indicates the 10% MTF.
Fig. 8.18. The sensor MTF of the CZT detector calculated as a function of thickness.

Fig. 8.19. The sensor MTF of the GaAs detector calculated as a function of donor concentration.

8.3.4 NPS

Fig. 8.20 shows the simulated 1-D NNPS of the CZT and the other semiconductors used here obtained at 10 μGy detector entrance air kerma. Fig. 8.21 demonstrates the dependence of the NNPS of the detector on the dose level.
Fig. 8.20 shows the NNPS of the CZT, Si and GaAs detectors calculated at 10 μGy entrance air kerma. Note that the 1-D NNPS was obtained by averaging the 2-D NNPS with a standard deviation of ±5.8%, on average.

Fig. 8.21 shows the NNPS of the CZT detector calculated as a function of incident air kerma. The relative error of the simulation is 5.8%, on average.
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8.3.5 DQE

The DQE of the CZT detector was calculated at 10 μGy detector entrance air kerma and is shown in Fig.8.22. Fig.8.23 shows the DQE calculated at 10 μGy detector entrance air kerma for all semiconductor detectors used here. Fig.8.24 shows the DQE of the CZT detector computed at three different levels of dose.

Fig.8.22. The DQE of the CZT detector calculated at 10 μGy detector entrance air kerma

Fig.8.23. The DQE of the CZT, CdTe, Si, GaAs and PbI₂ detectors calculated at 10 μGy detector entrance air kerma
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8.4 Discussion

8.4.1 Detector Characterisations

In addition to the active volume, the intrinsic detection efficiency of a detector depends on the material linear attenuation coefficient of that detector [5]. It can be seen from the plot of the linear attenuation coefficient of the CZT detector (Fig.8.10) that the photoelectric absorption is the dominant interaction for photon energy up to 100 keV. This means that the ratio of the photoelectric to Compton cross-section is very large in the CZT detector which would provide good quantum efficiency at the X-ray energy range of mammography.

It was also shown in Fig.8.11 that the CZT detector has an excellent quantum efficiency in the mammographic X-ray energy range. This is mainly due to the higher effective atomic number and density of the CZT detector which provided greater than 98% quantum efficiency at 20 keV for the 0.5 mm crystal thickness. At the same energy and thickness this was noted to be 64% for the Si detector. The CZT detector has also higher quantum efficiency than that obtained using the CsI (TI) which is the typical phosphor material used in the currently available indirect digital mammography systems.
The CZT detector has also excellent energy resolution in the range of mammography energy. It is shown in Fig.8.12 that the detector has a FWHM of 0.4 keV and 1.15 keV at 5.89 keV and 59.53 keV, respectively. It was also shown in Fig.8.12 that the experimental and simulation energy spectra are in good agreement which validates the rest of the results based on the simulation work.

8.4.2 Detector Response

Before measuring the MTF and NPS of any imaging detector, the detector response has to be linear with dose incident on the detector. If this is not the case (as it is in CR systems), the MTF and NPS images have to be linearised after the acquisition. However, the linearity of the CZT detector was verified and it was found that the detector has a linear relationship with dose, as shown in Fig.8.13. In fact, this type of relationship was reported previously [8] for all direct conversion digital detectors. This gives an advantage in using digital mammography over S/F systems which have a logarithmic relationship and hence lower dynamic range.

8.4.3 MTF

The sensor MTF of the CZT detector was found to be 40% at spatial frequency of 5 lp/mm. This means that the CZT detector can distinguish between two small objects 100 μm apart. The performance of the CZT detector was compared to the other semiconductor detectors in terms of sensor MTF instead of preMTF. It was found that the PbI₂ and GaAs detectors impart the highest MTF. While the CZT and CdTe detectors are comparable, the Si detector shows the lowest values of MTF.

The limited level (10%) of the sensor MTF of the Si detector was reached at spatial frequency of about 5 lp/mm. This would be much worse if a 100 μm pixel size, in which the Nyquist frequency is 5 lp/mm, is employed. This would suggest possible limitations with such a sensor in resolving objects close to the Nyquist frequency. The explanation of this poor resolution of the Si detector is the so-called K-photon reabsorption effect. Due to the low atomic number of Si, the photoelectric absorption is usually followed by a fluorescent kα photon with very low energy (1.74 keV) and short mean free path (12 μm). The probability of this photon to escape the detector is then very small and it would be instead reabsorbed close to where the primary
interaction occurred. This would result in expanding the initial charge cloud size which means broadening the single-event distribution within the sensor material.

As expected CZT and CdTe detectors have a comparable performance and this is because they have very similar detector properties. Their MTF is degraded by the low charge collection efficiency (CCE) due to the lower hole mobility of these detectors. PbI$_2$ and GaAs in turn show the highest values of MTF. While the probability of $K$-photon escape in a GaAs detector is rather large due to the relatively high energy (12.7 keV) that its characteristic photons would have, the $K$-edge of a PbI$_2$ detector is beyond the energy range of interest in this work. This means that there will be no $K$-fluorescence in the PbI$_2$ detector at mammographic X-ray energies.

Thus far, the cause of the degradation of the sensor MTF in a semiconductor detector is primarily due to the lateral spread of charge generated in the detector active volume. This lateral diffusion is influenced by several parameters. The most important ones are the thickness of the sensor and the dopant concentration. These two factors were investigated using the CZT and GaAs detectors and it was found that the sensor MTF is improved as donor concentration increases. Adding enough impurities to the intrinsic semiconductor material will increase the magnitude of the electric field in the fully depleted semiconductor (Eq.8.7). This in turn, increases the drift velocity of the carriers as it is in the linear region as a function of the electric field in the semiconductor (Eq.8.8). This means that the time required to the carriers to drift to the collection electrode decreases which would improve the charge collection efficiency (CCE) of such detector. However, this phenomenon is valid up to a certain amount of impurities concentration at which a saturation point is reached. This could be a result of the high density of trapping centres present at this point limiting the hole mobility – lifetime products in the compound semiconductor materials. Nevertheless, typical impurities concentration was reported previously [16] to be less than $5 \times 10^{12}$ cm$^{-3}$. This exactly agrees with the result presented here which is indicated by the area between the green and the red curves in Fig.8.19.

In scintillator detectors, the thickness of the sensor is a crucial factor due to the trade-off between the efficiency and the spatial resolution. The thicker the sensor, the higher is the quantum efficiency of the system. Spatial resolution, on the other hand is degraded due to the extreme lateral diffusion of the light photons emitted by such
thick phosphor. In contrast, semiconductors can be made with thickness sufficiently large to permit full absorption of the incident X-ray photons, but small enough so that their spatial resolutions are not affected. It was shown in Fig.8.18 that the CZT sensor can maintain its MTF in the range of thickness from 0.5 mm to just about 5 mm after which the MTF tends to drop significantly. This can be explained because as the detector thickness increases the more carriers are generated at depth far away from the collection electrode, thereby reducing the total CCE of the system. Nonetheless, a 0.5 mm thick CZT detector is sufficient to stop most of the photons in the 20 keV X-ray beam giving quantum efficiency that is greater than what one can expect from Si detector of 1 mm thickness [2]. Fig.8.11 showed that the 0.5 mm thick CZT detector provided greater than 98 % quantum efficiency while the Si detector recorded only 64% with the same thickness.

8.4.4 NPS

As expected for the direct conversion X-ray imaging system, the 1-D NNPS was found to be relatively constant across all frequencies. It was also found (the results are not shown) that all the detectors used here have an identical NPS (the absolute noise) but different NNPS (noise relative to mean pixel value), as shown in Fig.8.20. While the Si detector has significantly the highest NNPS, the other detectors showed identical NNPS with minute differences. This was noted to be 0.000170, 0.000110, 0.00012, 0.00010, 0.00011 (±5.8%) for Si, CZT, CdTe, GaAs and PbI₂ at spatial frequency of 2.5 lp/mm. This can be explained as the Si has lower atomic number than those for the other detectors and hence the lower mean pixel value.

The results have shown that the NNPS of these detectors strongly depends on the exposure level. It was noticeable that the relative noise (NNPS) decreased with increasing dose. In other words, NNPS is proportional to the mean pixel value which in turn has a linear relationship with dose incident on the detector.

8.4.5 DQE

The DQE of the semiconductor detectors used here was calculated for 10 μGy incident air kerma and plotted as a function of spatial frequency. It was clearly seen that the Si detector showed the lowest DQE among all other detectors. While the CZT and CdTe detectors have an identical DQE, GaAs and PbI₂ detectors exhibited the
highest values of DQE. The DQE values were reported to be 0.48, 0.46, 0.56 and 0.54 (±5%) at spatial frequency of 1 lp/mm and 0.25, 0.22, 0.31 and 0.30 (±5%) at spatial frequency of 2.5 lp/mm for CZT, CdTe, GaAs and PbI2, respectively.

The exposure dependence of the DQE of the CZT detector was investigated by calculating the DQE of the detector at five different exposure levels. It comes as no surprise that the DQE has no change with dose. This can be explained as the noise simulated here represents only the quantum noise. In practice, DQE would otherwise increase with dose due to the presence of electronic noise. The DQE, however, would increase up to a certain level of dose at which the electronic noise becomes negligible and the detector is considered as quantum limited noise.

8.5 Conclusions

In this chapter, MTF, NNPS and DQE of selected direct conversion detectors were measured with the aid of MC technique. The results have shown that the PbI2 and GaAs detectors imparted the highest MTF over all other semiconductors detectors investigated in this work. While the CZT and CdTe were comparable, Si detector showed the lowest values of MTF.

NNPS of these detectors was also studied and found to be identical and relatively uniform across all spatial frequencies. The results demonstrated a strong dependence of the NNPS on dose. In other words, NNPS is proportional to the mean pixel value which in turn has a linear relationship with entrance air kerma. This could explain the independence of the DQE on the level of dose especially in the absence of electronic noise. Nevertheless, the results have shown that the PbI2 and GaAs have the highest DQE with little difference between the two detectors, the CZT and CdTe were similar and the Si detector showed the lowest values of DQE at all spatial frequencies.

Finally, the distribution of X-ray photons interacting in a semiconductor material and the lateral diffusion of charges generated in that material were modelled with the aid of the MC and analytical methods. The model was applied to digital mammography in order to investigate the characteristics of some semiconductor detectors and study the factors which degrade their spatial resolution. The study of the sensor MTF of a detector away from its aperture function is a useful way to improve the spatial

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resolution of that detector without changing the pixel size. Effect of charge carrier mobility and dopant concentration on the spatial resolution of semiconductor detectors was investigated and significant relationship was found.

References


Chapter 9

Conclusion and Future Work

The applicability of the current UK and European assessment procedures was investigated for digital mammography systems. These include the procedures for measuring image quality in terms of contrast detail detectability and CNR analysis. The impact of the ROI size on the relative noise and CNR measurements was investigated for CR and DR systems. It was found that the measured relative noise for the CR images strongly depended on the ROI size due to the heel effect. In this case the heel effect distorted the CNR measurement when the ROI specified in the UK and European guidelines was used. After applying heel effect correction there was very little dependence on ROI size. This dependence did not arise when testing DR systems, presumably due to the flat-field correction applied by the manufacturer. However the use of multiple very small ROIs led to a result that was essentially the same as if a heel effect correction had been applied. One more advantage of this approach is that only a single image is required, while the application of a heel effect correction requires two images, for CNR and relative noise determination of CR systems.

The automated approach of reading CDMAM images was further investigated and compared to the human readings. The reproducibility of automatic reading and its ratio to the human observer was then determined using a large number of human readers and a large population of digital mammograms for different types of systems. It was found that the automated measurements can be used to predict the threshold contrast for a typical observer. Despite some limitations automated reading of CDMAM images can provide a reproducible means of assessing digital mammography systems against European Guidelines. However, using larger numbers of images and a change in phantom design could greatly improve the reproducibility of the automatic observer.
CHAPTER 9. CONCLUSION AND FUTURE WORK

The predicted human contrast detail measurements were then used to evaluate and compare image quality of a wide range of CR and DR digital mammography systems. The contrast detail detectability of these systems was determined and analysed across a range of doses and the results were presented as the dose required to achieve the standards in the European guidelines at different detail sizes. It was found that the dose required to meet the image quality standards for the indirect DR system was rather constant for all detail sizes of the CDMAM phantom. The direct DR systems, on the other hand, seemed to need slightly lower dose levels for the smaller details than those required for the larger details. This may be explained by the relatively good MTF for the a-Se technology used by these systems. The dose required to meet the image quality standards for the CR systems was found to be very dependent on the detail size considered. It was also observed that for most CR systems the achievable image quality cannot be reached for doses less than the upper acceptable limit in the guidelines of 3 mGy. The fact that dose requirements are higher for CR systems for small details is presumably due to their relatively low MTF at higher spatial frequencies. The Konica CR system was tested with three different plates with different MTFs. The plate with the best MTF and DQE (CP-1M), also required the lowest dose for detecting the smallest details.

However, despite the feasibility of using the automatic observer to predict contrast detail measurements, this approach suffers from two main limitations. One is that such a method requires a large number of images to reduce statistical errors. The other limitation is that using the automatic method produces results that are different from those of human observers. An alternative approach would be to model contrast detail response from objective measurements such as MTF and DQE. Thus, MTF NNPS, NEQ and DQE were obtained for the detectors used in a wide range of currently available digital mammography systems and the results compared. It was found that the DR systems, in general, have relative uniform NNPS curves over a large spatial frequency band. This is presumably a result of their MTF which remains high up to the Nyquist frequency. The MTF and NNPS of these systems were found to be reasonably radial isotropic. While the NNPS of the photon counting system was symmetric in both directions, the MTF was significantly lower in the slit scan direction than that in the array direction. This was expected as the MTF of a scanning multi-slit system in the array direction is detruled by the strip pitch, whereas, in the
scan direction, it is defined by the precollimator slit width and the continuous scan motion. The CR systems showed significant asymmetric MTF and NNPS between the scan and subscan directions. This is likely due to differences in signal processing in the two directions.

The DQE of the DR systems, generally, increased with dose until a plateau is reached. The DQE fall-off at low dose is possibly due to the effect of the electronic noise which becomes pronounceable relative to the quantum noise. The DQE of the CR systems, in contrast, decreases with an increase in X-ray dose, due to the presence of the large amount of fixed-pattern noise which is proportional to the exposure. However, the use of the crystal needle CR plate would improve the DQE of the CR system, at the clinical levels of exposure.

In general, a strong relationship was found between the dose required to meet the acceptable requirements in European Guidelines and the DQE of the detector systems. Thus a relatively large dose required to meet the image quality limits is expected for a system with a relatively low DQE. In practice for a given system this relationship will depend on some other factors such as the scatter rejection method and beam quality. However, where a manufacturer has achieved improvements in DQE this is reflected in lower doses being needed to meet image quality standards in European Guidelines. Nevertheless, under the current guidance a very wide range in performance, in term of dose and image quality, is accepted and used clinically.

Subsequently, a simple signal-matched noise-integration model was then applied to theoretically predict the contrast detail response of the systems examined. In this model the MTF and DQE of the detectors of these systems were used to model the contrast detail measurements. Using this model an encouragingly good level of agreement was found between the experimental data and theoretical predictions.

Despite the broadly reasonable contrast detail measurements predicted by the model, the model suffers from a number of simplifications compared to the experimental methods. These are
 CHAPTER 9. CONCLUSION AND FUTURE WORK

a) The model assumes that the detector entrance exposure (Xe) is the same for all detail sizes. In the experimental methods the exposure varies across the detector and will correlate systematically with detail size due to the design of the phantom.

b) A very simple approach has been taken to estimate the contrast reducing effect of scatter by assuming that it is the same for all detail sizes. In the experimental method the amount of scatter is likely to vary with position across the image of the phantom and may therefore vary with detail size.

c) The model uses only the zero frequency DQE and therefore takes only a very simple approach to accounting for the frequency response of the system.

d) The model does not take account of the effect of pixel size approaching the size of the details being detected.

Therefore, further development of the model is suggested for future work. This development should take into consideration all the limitations mentioned above. A more refined version of this model would have the potential to predict contrast detail performance from standard measurements of detector performance and may provide a more reproducible means of comparing the performance of digital mammography systems.

There is ongoing research on direct conversion hybrid pixel semiconductor detectors for the application to digital mammography. Therefore, evaluation of image performance of a variety of these detectors was one of the aims of this thesis. MTF, NNPS and DQE were calculated with the aid of MC simulation and theoretical techniques. The results have shown that the PbI₂ and GaAs detectors exhibited the highest MTF and DQE over all other semiconductors detectors investigated. It was found that in the absence of added noise the DQE of these detectors would have become dose independent.

The model has shown that the study of the sensor MTF of a detector away from its aperture function is a useful approach to improve the spatial resolution of that detector without changing the pixel size. It has also shown that image performance of
compound semiconductor detectors strongly depended on charge carrier mobility and dopant concentration.

However, the model assumes ideal detectors, with 100% collection efficiency, which is not the case in reality. Therefore, a further exploration of the impact of the charge collection efficiency (CCE) on the MTF of the detectors is suggested for future work. The CCE is defined as the ratio of charge carriers collected at the electrodes to the total charge carriers generated by the X-ray interactions in the sensor material. The CCE of a semiconductor detector can be determined by means of the Hecht equation, as

\[
CCE(z) = \left( \frac{\lambda_e}{d} \right) \left( 1 - e^{-\frac{(d-z)}{\lambda_e}} \right) + \left( \frac{\lambda_h}{d} \right) \left( 1 - e^{-z/\lambda_h} \right)
\]

where \(z\) is the origin of charge carriers generated in the crystal with \(d\) thickness, and \(\lambda_e\) and \(\lambda_h\) are the mean free paths calculated as \(\mu_e \tau_e E\) and \(\mu_h \tau_h E\), where \(\mu_e \tau_e\) and \(\mu_h \tau_h\) are the mobility- life time products for electron and hole, respectively, and \(E\) is the electric field strength which is equal to \(V/d\) where \(V\) is the applied voltage in volts.
Appendices

Appendix A: Visual Basic code created for automated analysis of multiple CDCOM output matrices

Note that the multiple CDCOM output matrices were generated randomly using a programme written in C++ language. The programme was developed by Jon Denne, Computing Section, Medical Physics Department, Royal Surrey County Hospital, Guildford. The programme was solely generated for the purpose of the work produced in Chapter 5.

' Created on 18/04/2007 by Abdulaziz Alsager, University of surrey
' Purpose: analysis of multiple RunCDCOM outputs (multiple fractions)

Sub multiple_inputs()
'
    Dim thisname As String
    thisname = ActiveWorkbook.Name
    Application.ScreenUpdating = False
    Workbooks.Open (ActiveWorkbook.Path & "\" & "fraction.csv")
    'opens runcdcom output file from same directory as spreadsheet
    Windows("fraction.csv").Activate
    Range("A2:P2700").Select
    Selection.Copy
    Windows(thisname).Activate
    Sheets("Multiple inputs").Select
    Application.ScreenUpdating = False
    Range("A2:P2700").Select
    Selection.PasteSpecial Paste:=xlPasteValues, Operation:=xlNone,
    SkipBlanks:=False, Transpose:=False
    Range("A1").Select
    Windows("fraction.csv").Activate
    Application.DisplayAlerts = False 'turns off alert to avoid dialog box
    Workbooks("fraction.csv").Close SaveChanges:=False
    Application.ScreenUpdating = True
End Sub

Sub Multiple_analysis()
'
    For batch = 1 To 150 'loop for each fraction
        Sheets("multiple inputs").Select
        Application.ScreenUpdating = False 'turn off the sheet while running the macro
        i = (batch - 1) * 18 + 3
        j = batch * 18
        Range("A" & i & ";P" & j).Select 'selecting the range of each fraction according to values of i and j
        Selection.Copy 'copy the fraction
        Sheets("input data").Select
        Application.ScreenUpdating = False
Range("B14:Q29").Select
Selection.PasteSpecial Paste:=xlPasteValues, Operation:=xlNone,
SkipBlanks:=False, Transpose:=False  " paste the fraction in the
input data sheet for the analysis
Application.ScreenUpdating = False
Application.Run "Master"
Application.ScreenUpdating = False
Sheets("export").Select
Application.ScreenUpdating = False
Range("H10:H21").Select
Selection.Copy  " copy unsmooth predicted gold output
Sheets("Multiple outputs").Select
Cells(2, 1 + batch).Select
Selection.PasteSpecial Paste:=xlPasteValues, Operation:=xlNone,
SkipBlanks:=False, Transpose:=False  " paste the output
Sheets("export").Select
Range("I10:I21").Select
Selection.Copy  " copy unsmooth fit to predicted output
Sheets("Multiple outputs").Select
Cells(15, 1 + batch).Select
Selection.PasteSpecial Paste:=xlPasteValues, Operation:=xlNone,
SkipBlanks:=False, Transpose:=False  " paste the output
Sheets("export").Select
Range("M10:M21").Select
Selection.Copy  " copy smooth predicted gold output
Sheets("Multiple outputs").Select
Cells(28, 1 + batch).Select
Selection.PasteSpecial Paste:=xlPasteValues, Operation:=xlNone,
SkipBlanks:=False, Transpose:=False  " paste the output
Sheets("export").Select
Range("N10:N21").Select
Selection.Copy  " copy smooth fit to predicted gold output
Sheets("Multiple outputs").Select
Cells(41, 1 + batch).Select
Selection.PasteSpecial Paste:=xlPasteValues, Operation:=xlNone,
SkipBlanks:=False, Transpose:=False  " paste the output
Next batch  " go to the next fraction
Application.ScreenUpdating = True
Range("EV2").Select
Application.CutCopyMode = False
ActiveCell.FormulaR1C1 = "=(AVERAGE(RC[-150]:RC[-1]))"  "
average the unsmooth predicted output
Range("EV2").Select
Selection.AutoFill Destination:=Range("EV2:EV13"),
Type:=xlFillDefault

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Range("EV2:EV13").Select
Range("EW2").Select
ActiveCell.FormulaR1C1 = "=(STDEV(RC[-151]:RC[-2]))"  ''
standard deviation of the unsmooth predicted output
Range("EW2").Select
Selection.AutoFill  Destination:=Range("EW2:EW13"),
Type:=xlFillDefault
Range("EW2:EW13").Select
Range("EV15").Select
Application.CutCopyMode = False
ActiveCell.FormulaR1C1 = "=(AVERAGE(RC[-150]:RC[-1]))"  ''
average the unsmooth fit to predicted output
Range("EV15").Select
Selection.AutoFill  Destination:=Range("EV15:EV26"),
Type:=xlFillDefault
Range("EV15:EV26").Select
Range("EW15").Select
ActiveCell.FormulaR1C1 = "=(STDEV(RC[-151]:RC[-2]))"  ''
standard deviation of the unsmooth fit to predicted output
Range("EW15").Select
Selection.AutoFill  Destination:=Range("EW15:EW26"),
Type:=xlFillDefault
Range("EW15:EW26").Select
Application.ScreenUpdating = True
Range("EV28").Select
Application.CutCopyMode = False
ActiveCell.FormulaR1C1 = "=(AVERAGE(RC[-150]:RC[-1]))"  ''
average the smooth predicted output
Range("EV28").Select
Selection.AutoFill  Destination:=Range("EV28:EV39"),
Type:=xlFillDefault
Range("EV28:EV39").Select
Range("EW28").Select
ActiveCell.FormulaR1C1 = "=(STDEV(RC[-151]:RC[-2]))"  ''
standard deviation of the smooth predicted output
Range("EW28").Select
Selection.AutoFill  Destination:=Range("EW28:EW39"),
Type:=xlFillDefault
Range("EW28:EW39").Select
Application.ScreenUpdating = True
Range("EV41").Select
Application.CutCopyMode = False
ActiveCell.FormulaR1C1 = "=(AVERAGE(RC[-150]:RC[-1]))"  ''
average the smooth fit to predicted output
Range("EV41").Select
Selection.AutoFill  Destination:=Range("EV41:EV52"),
Type:=xlFillDefault
Range("EV41:EV52").Select
Range("EW41").Select
ActiveCell.FormulaR1C1 = "=(STDEV(RC[-151]:RC[-2]))"  ''
standard deviation of the the smooth fit to predicted output
Range("EW41").Select
Selection.AutoFill  Destination:=Range("EW41:EW52"),
Type:=xlFillDefault
Sub DeleteMultipleOutputs()
    Range("B2:EW13").Select
    Selection.ClearContents
    Range("B15:EW26").Select
    Selection.ClearContents
    Range("B28:EW39").Select
    Selection.ClearContents
    Range("B41:EW52").Select
    Selection.ClearContents
End Sub
Appendix B: MCNPX file created for the simulation work

c title Modelling Line Source Tilted 3 degree
  Planer CTE divided into pixels
  X-ray spectrum source (Mo with 28 kV, 0.128mm Be window and 0.03mm Mo filtration)
  MCNPX code created by Abdulaziz Alsager, University of surrey

c cell card
  1 0 -100 fill=1 IMP:p 2
  2 1 -6.0 -101 lat=1 u=1 IMP:p 2
  1 3 2 -1.29e-3 #1 #2 -105 IMP:p 1
  sphere
  4 0 105 IMP:p 0

c end of cell card

c surface card
  100 rpp -2.21 2.21 2.21 2.21 0 0.05
  101 rpp -0.00425 0.00425 -0.00425 0.00425 0.00425 0.00425 0 0.05
  105 SO 140

c end of surface card

c data card
  SDEF X=d1 Y=d2 Z=0.051 par=2 erg=d4 vec= 0 0 -1 TR=1
  SI4 0.5 0.5
  SP1 0 1
  SP2 0 1
  # SI4 SP4
  0.0005 0.00E+00 L D
  0.001 0.00E+00
  0.0015 0.00E+00
  0.002 0.00E+00
  0.0025 0.00E+00
  0.003 0.00E+00
  0.0035 4.46153E-20
  0.004 9.76782E-15
  0.0045 2.97166E-11
  0.005 6.91017E-09
  0.0055 3.33151E-07
  0.006 5.39646E-06
  0.0065 4.22748E-05
  0.007 0.000196602
  0.0075 0.000634287
  0.008 0.001514844
  0.0085 0.003157302
  0.009 0.005430322
  0.0095 0.008456188
  0.01 0.012090495
  0.0105 0.015913960
  0.011 0.019878985
  0.0115 0.023682995
  0.012 0.027338663
  0.0125 0.030269554
  0.013 0.032999752
  0.0135 0.035103713
  0.014 0.036753465
  0.0145 0.038011386

200
0.015 0.038940594
0.0155 0.039261881
0.016 0.039461386
0.0165 0.039282426
0.017 0.03882797
0.0175 0.284589109
0.018 0.037042822
0.0185 0.035811386
0.019 0.034561139
0.0195 0.076881683
0.020 0.03588416
0.0205 0.0032323837
0.021 0.003241559
0.0215 0.003222723
0.022 0.003233342
0.0225 0.003269752
0.023 0.003288193
0.0235 0.003292772
0.024 0.003247376
0.0245 0.003157772
0.025 0.00297755
0.0255 0.00273094
0.026 0.002399564
0.0265 0.001945921
0.027 0.001401941
0.0275 0.000750057
0.028 6.69733E-05

C SI5 0 0.9763 1.0
C SP5 0 1.0 1.0

*TR1 0 0 0 3 -87 90 93 90 90 90 0

Mode p

*F8:p (2<2[-26:25 -26:25 0])

C Material cards
C Detector material
M1
30000 -0.093671596 $ Zn
48000 -0.37468384 $ Cd
52000 -0.531642020 $ Te
M2
006000 -0.000124 $ Air
007000 -0.755267
008000 -0.231781
012000 -0.012827

NPS 1.0E8
PRDM 0 0 1

Print
Appendix C: Matlab code created for randomly blurring charge clusters at the detector surface

Programme Name: Psf_image_blurring: created by Abdulaziz Alsager, University of Surrey

Purpose: this Programme Blurring an unblurred image by sigma

Input:
(1) MCNPX file
(2) dimension of the detector
Output: Image that spatially blurred

clear all; close all;
input_file = input ('Enter the name of the image file: ','s') ;
sigma = input('Enter the value of positional sigma in cm: ');
pixel_size = input('Enter the pixel size in cm: '); 
sigma_in_pixels = sigma/pixel_size;
IMAGE = load(input_file);
Factor=10;
IMAGE=Factor*IMAGE;

image_size = length(IMAGE);
NEW_IMAGE = zeros(image_size,image_size);
for i = 1:image_size;
    for j = 1:image_size;
        count = IMAGE(i,j);
        if(count>0);
            for k=1:fix(count);
                a=fix(i+randn*sigma_in_pixels);
                b=fix(j+randn*sigma_in_pixels);
                if(a>=1 & a<=image_size & b>=1 & b<=image_size);
                    NEW_IMAGE(a,b) = NEW_IMAGE(a,b) + 1 ;
                end;
            end;
        end;
end;
figure(1);
subplot(2,1,1);imshow(IMAGE, []); title('Old Image') ;
subplot(2,1,2);imshow(NEW_IMAGE, []); title('Blurred Image') ;
figure(3)
mesh(IMAGE);
xlabel('Distance (Pixels)'); ylabel('PSF Value');
title('The original 3D Plot') ;
figure(4)
mesh(NEW_IMAGE);
xlabel('Distance (Pixels)'); ylabel('PSF Value');
title('The blurred 3D Plot');
Appendix D: List of Publications, conferences and workshop

A) Publications

I. Scientific Papers


4. Abdulaziz A. Alsager, Jennifer M. Oduko, Ozcan Gundogdu, Kenneth C. Young and Nicholas M. Spyrou, “Evaluation of modulation transfer function measurements in digital mammography”, Accepted to be published in the Journal of Medical Physics, March 1, 2009


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**II. Technical Reports**


**III. MSc dissertations (co-supervision)**


B) Conferences and Workshops

1. Participated in a workshop on the Monte Carlo Simulation Technique held on 27th - 31st of March 2006 at iThemba Lab in Cape Town, South Africa
2. Participated and gave a scientific oral presentation in University Nuclear Technology Forum (UNTF) Conference, University of Sheffield, Sheffield, UK, April 2006
3. Participated and presented a scientific poster in International Symposium on 10th Radiation Physics (ISRP-10), University of Coimbra, Coimbra, Portugal, 17th - 22nd, September 2006
5. Participated and gave a scientific oral presentation, EUROCON 2007 – The International Conference on “Computer as a Tool”, Warsaw, Poland, 9th - 12th September 2007
6. Presented a poster, Workshop on Uncertainty Assessment in Computational Dosimetry, Bologna, Italy, 8th - 10th October 2007
7. Participated and gave a scientific oral presentation 27th Meeting of the UK Mammography Physics Group (UKMPG), St. Mary’s Church, Sheffield, 1st November 2007