

The copyright of this thesis vests in the author. No quotation from it or information derived from it is to be published without full acknowledgement of the source. The thesis is to be used for private study or non-commercial research purposes only.

Published by the University of Cape Town (UCT) in terms of the non-exclusive license granted to UCT by the author.

Ph.D. Thesis

Computer models for the design and
optimization of a linear slot scanning
mammography system

Khalid Ibrahim Hussein

Supervisor: Assoc. Prof. Tania Douglas

Co-supervisor: Prof. Christopher L. Vaughan

Submitted to the Faculty of Health Sciences at
the University of Cape Town in fulfillment of the requirements for the degree of
Doctor of Philosophy in Biomedical Engineering
Cape Town, South Africa, June 2008

Declaration

Computer models for the design and optimization of a linear slot scanning mammography system

I, Khalid Ibrahim Hussein, hereby declare that:

- (a) the above thesis is my own work both in concept and execution, and that apart from the normal guidance from my supervisors, I have received no assistance except that stated in the acknowledgements.
- (b) except as stated below, neither the whole nor any part of the above thesis has been submitted in the past, or is being, or is to be submitted for a degree at the University of Cape Town or any other university.

The thesis has been presented by me for examination for the degree of Doctor of Philosophy in Biomedical Engineering.

Signature

Date

Acknowledgements

I would like to express my sincere gratitude to the following people for their assistance and guidance:

Supervision: A big thank you to Dr. Tania Douglas for her support and encouragement towards both academic and personal growth; to my second supervisor Prof. Christopher L. Vaughan for his useful guidance and support; and to Dr. Julien Marchal for guidance and helpful ideas during the first stage of the project.

Data collection: My thanks to the Radiography departments at Groote Schuur, Red Cross, Panorama Medi-Clinic, and Mitchells Plain hospitals for their flexibility and patience during the collection of my data.

Mammography group: My sincerely grateful to thanks the mammography team: Prof. Kit Vaughan, Dr. Tania Douglas, Mr. Stef Steiner, Dr. Peter Gresak, Mr. Michael Evans, Mr. Charles Harris and Ms. Virginia Sanders. My thanks to Stef Steiner and Charles Harris for their help and support in the mechanical design and assembly of the mammography system; to Peter Gresak for his patience, good nature, and guidance, and for helping me in the collection of some of the data; and to Michael Evans for his suggestions, help, support, and outstanding programming skills.

My Family: Finally, I would like to thank my wonderful family, my parents, my wife Maie, and my two cute sons Mohammed and Omer for everything they have done for me and for their patience and encouragement during my studies.

This PhD project formed part of an NIH-funded project (5R21CA101705-02) “Low dose digital X-rays for mammographic screening” conducted in the MRC/UCT Medical Imaging Research Unit, University of Cape Town.

Abstract

Over the past years, digital x-ray mammography has developed into a powerful imaging tool for breast cancer screening. In spite of the limitations in photon flux and tube loading, slit or slot scanning configurations are considered to be solutions for better image quality. Linear slot scanning radiography (LSSR) using a narrow fan beam has shown a significantly lower dose than that of conventional x-ray systems. Lodox Statscan is a full body linear slot scanning radiography system approved by the US Food and Drug Administration (FDA). The relatively high resolution and significantly lower dose of the Lodox Statscan system gives rise to the question: *Can the linear slit or slot scanning configuration used in the Lodox system be used in mammography to produce image quality similar to or better than that of current digital systems with corresponding low dose?* The purpose of this thesis is to answer this question.

Presented here, is the design and optimization of a linear slot scanning mammography system based on the Lodox Statscan system. The design specifications and system requirements were determined using computational mathematical tools developed during the project. The system was designed to have a LSSR configuration with a mammography x-ray tube and a high performance detector providing the required power and resolution for mammography. The system was evaluated using a tissue-equivalent physical phantom designed for mammography, in order to compare its performance with that of existing mammography systems and to establish image quality parameters such as modulation transfer function, noise power spectrum, and detective quantum efficiency. This evaluation showed the ability of the system to detect microcalcifications (high contrast objects) of 0.130mm and hemispheric masses (low contrast objects) of 1.19mm at mean glandular dose of 0.5mGy with a good resolution of nearly 9lp/mm. Further assessment of the system showed a detector linearity response of 0.9994 for entrance exposures over the range 0.3 to 3.0 mGy. The system additive noise measured using dark images showed a value of 4000 electrons, which is considered high compared to current mammography systems. The mammography system showed a detective quantum efficiency of 0.56 at mean glandular dose of 0.5mGy, which is comparable to that of existing digital mammography systems that use the indirect detection method, but is lower than that of systems using direct and photon counting detectors.

A mathematical tissue-equivalent phantom was developed for mammography system optimization. The mathematical phantom was evaluated using the mammography system and the physical tissue-equivalent phantom. The results obtained using the mathematical phantom were in good agreement with values measured using the physical tissue-equivalent phantom. The estimated pixel intensity of the mathematical phantom, the analogue-to-digital conversion gain, and the detector additive noise showed good agreement with measured values, with a correlation approaching 1. An application of the model, to examine the feasibility of using a monochromatic filter for dose reduction and improvement of image quality for slot scanning digital mammography, is also presented. A significant mean glandular dose reduction was found using a monochromatic beam compared to a polychromatic beam. The monochromatic beam is combined with a three-fold increase in tube load to achieve the same signal-to-noise ratio as a polychromatic beam; this factor can be reduced by using a wider monochromatic beam.

A mathematical tool was developed to study the effects of detection thresholds and the anode-filter combination on the image quality when using a photon counting detector. Results showed that the broad-spectrum detective quantum efficiency (DQE) and figure of merit (FOM) degrade when increasing the detection threshold. Using the W:Al anode-filter combination rather than Mo:Mo provided a smoother DQE curve especially for larger breast thicknesses, due to the absence of sharp peaks in the W:Al spectrum.

Despite the fact that slot scanning has a limitation regarding the photon flux, the improvement in image contrast and the reduction in dose have an impact on image quality, radiation hazard, and patient comfort. This makes linear slot scanning a promising technique for digital mammography.

Table of Contents

Declaration	i
Acknowledgements	ii
Abstract	iii
Table of Contents	vi
List of Figures	xi
List of Tables	xvii
1 Introduction	1
1.1 Project purpose and scope	1
1.2 Document overview	3
2 Background Theory	6
2.1 Anatomy of the breast	6
2.2 Screening for breast cancer	7
2.2.1 Screen-film mammography	7
2.2.2 Digital mammography	7
2.3 Image quality parameters	10
2.3.1 Signal-to-noise ratio (SNR)	10
2.3.2 Linear Cascaded Model	13
2.3.3 SNR and DQE	20

2.4	Lodox Statscan	20
3	System Requirements for Linear Slot Scanning Mammography	22
3.1	Introduction	22
3.2	System design and requirements	23
3.2.1	Mechanical design	23
3.2.2	X-ray tube	24
3.2.3	Detector specifications	26
3.2.4	Time delay integration (TDI) mode	30
3.2.5	Summary of system specifications	32
3.3	Modelling the performance of LSSM	32
3.3.1	Energy spectrum simulation	32
3.3.2	Imaging task	35
3.3.3	Scatter	35
3.3.4	System resolution MTF	39
3.3.5	Noise power Spectrum NPS	39
3.3.6	Detective quantum efficiency DQE	41
3.4	Summary and discussion	43
3.5	Author's contribution	44
4	Linear Slot Scanning Mammography System Evaluation	45
4.1	System description and specifications	46
4.1.1	System description	46
4.1.2	System specifications	46
4.1.3	X-ray source and collimation	47
4.1.4	LSSM detector	47
4.1.5	Operator user interface and image acquisition	48
4.2	System evaluation	48
4.2.1	Spectrum calibration and dose measurements	49

4.2.2	Beam quality measurement	51
4.2.3	Detector response	52
4.2.4	Image quality measurements	53
4.2.5	System evaluation using a breast tissue-equivalent phantom	58
4.3	System Comparison Summary	64
4.4	Summary and discussion	65
4.5	Author's contribution	66
5	Modelling and Validation of a Mathematical Tissue-Equivalent Breast Phantom for Digital Mammography	67
5.1	Introduction	67
5.2	Methods	68
5.2.1	Modelling the pixel intensity in digital units	70
5.2.2	Estimation of the image contrast	71
5.2.3	Modelling the detector noise	72
5.2.4	Modelling the total system resolution	73
5.3	Generation of phantom image	75
5.4	Validation of the mathematical phantom	76
5.4.1	Estimation of digital pixel intensity and ADC conversion gain	76
5.4.2	Estimation of the image contrast	80
5.4.3	Estimation of contrast-to-noise ratio	81
5.4.4	Summary and discussion	84
5.5	Author's contribution	86
6	Applications of the Mathematical Breast Phantom	87
6.1	Introduction	87
6.2	Evaluating different detector configurations	88
6.2.1	Method	88
6.2.2	Results	89
6.3	X-ray beam optimization	91

6.3.1	Background	91
6.3.2	Method	92
6.3.3	Results	94
6.4	Summary and discussion	97
6.5	Author's contribution	98
7	Photon Counting Threshold Optimization in Mammography	99
7.1	Introduction	99
7.2	Single photon counting and detection thresholds	101
7.3	Methods	102
7.3.1	X-ray spectra	102
7.3.2	Detector parameters	103
7.3.3	Imaging Task	103
7.3.4	Mean glandular dose and scatter-to-primary ratio	103
7.4	Results	104
7.5	Summary and discussion	109
7.6	Author's contribution	110
8	Conclusions and Recommendations	111
A	Slot Scanning Mammography: MATLAB Graphical User Interface	114
A.1	Introduction	114
A.2	Software Requirements	115
A.3	Basic Usage	115
A.4	MammoModel GUI	116
A.4.1	Generation of x-ray energy spectrum	118
A.4.2	Imaging Task	119
A.4.3	Detector Parameters	120
A.4.4	Output parameters	121
A.5	PhantomUser GUI	124

A.5.1	Image Processing	126
A.5.2	MXF Performance	127
A.6	Optimum Threshold GUI	128
A.6.1	Technique Factors	130
A.6.2	Imaging Task	130
A.6.3	Tube voltage	130
A.6.4	Threshold energy	130
A.6.5	Photon energy	130
B	The linear slot scanning mammography system components	132
B.1	C-arm and x-ray source	132
B.2	The collimation system	133
B.3	The detector and its components	133
C	Hamamatsu charge-coupled device (CCD) data sheet, April 2003	134
D	Data sheet: Tissue-Equivalent Phantom for Mammography	139
E	Matlab Codes	142
	References	143

List of Figures

2.1	Breast anatomy, adapted from (BreastCancer, 2007)	6
2.2	The detector components of a digital system (on the left) and their function (on the right).	8
2.3	The Rose model	11
2.4	Linear cascaded model	14
2.5	The Lodox Statscan machine located in MRC/UCT Medical Imaging Research Unit	21
3.1	(a) The slot scanning system; (b) geometry of the slot scanning digital mammography system; SDD is the source-to-detector distance, SCD is the source-to-collimator distance, SAD is the source-to-axis distance, C_W is the collimating slot, B_W is the beam width at the patient axis, and Ag is the airgap.	24
3.2	The limiting resolution as a function of focal spot size along the length (slot direction) and along the width (scan direction) of the scanning direction.	25
3.3	Varian tube limits.	26
3.4	The modulation transfer function (MTF) and absorption efficiency for a high light output (HL) CsI scintillator calculated using <i>MammoModel</i> and data provided by Hamamatsu Company (FOS, 1996); (a) the value of MTF at 10lp/mm frequency for each FOS scintillator thickness; (b) absorption efficiency as function of FOS scintillator thickness at 40kVp.	27
3.5	Coupling of the scintillator to the CCD (a) direct coupling, and (b) coupling through fibre optic taper (FOT), adapted from (FOS, 1996).	29
3.6	Schematic illustration of the TDI mode of operation.	30
3.7	The estimated maximum charge capacity for a direct and indirect coupling detectors using the Hamamatsu technical specifications and <i>MammoModel</i> described in Appendix A.	34

3.8	The calculated tungsten energy spectrum at a distance of 60cm, tube voltage of 30kVp, tube current of 100mA, and external filtration of 0.5mm Al.	34
3.9	The calculated energy spectrum modified for slot scanning; (a) the filtered spectrum (0.5mmAl, 6mm beam width, 35kVp and 200mA at 60cm from the source) and attenuated spectrum through 4cm breast (50:50 glandular/adipose tissue); (b) calculated spectrum for different imaging tasks, 50:50 breast and different microcalcifications (100, 400, and 700 μm	37
3.10	The scatter effect in slot scanning.	37
3.11	The effect of slot width in scatter rejection for different breast thickness.	38
3.12	The effect of slot width in scatter rejection for different kVp.	38
3.13	The estimated focal spot MTF using the measured intensity distribution of the Varian x-ray tube (RAD 70, 0.3mm focal spot)(RAD-70, 2004): (a) measured intensity distribution in direction along focal length (slot direction) and along width (scan direction); (b) simulated intensity distribution; (c) calculated MTF in slot and scan direction; (d) calculated focal spot MTF for different object magnifications (m), magnification M=2 means object midway between detector and source.	40
3.14	The scintillator MTF.	41
3.15	The estimated system MTF for LSSM system using a direct coupling detector consisting of the Hamamatsu FOS and CCDs.	41
3.16	The normalized noise power spectrum (NNPS) for the LSSM detector; data are shown for three detector exposures.	42
3.17	The normalized noise power spectrum (NNPS) for the LSSM detector; data are shown for different levels of electronic noise.	42
3.18	The detective quantum efficiency (DQE) as a function of scintillator thickness.	43
3.19	The detective quantum efficiency (DQE) as a function of coupling efficiency.	43
4.1	The LSSM system configuration.	46
4.2	Slot scanning mammography system (LSSM); numbered components are described in Table 4.1.	48
4.3	The calculated energy spectrum modified for slot scanning (a) different beam widths (BW), 35kVp and 200mA (b) different tube voltages, beam width 6mm and 200mA.	49
4.4	The mean glandular dose MGD as function of measured entrance dose at the surface of the phantom.	50

4.5	The estimated and measured entrance exposure at a distance of 60cm from the source at different tube voltages, with a correlation of 0.9999, and slope of 0.96 for 35kVp.	50
4.6	The measured entrance exposure as a function of distance from the x-ray source and the estimated dose using the inverse distance ($1/r$) and the inverse square distance ($1/r^2$) law.	51
4.7	The Half Value Layer of the LSSM x-ray beam using external filtration of (a) 0.5mm Al; (b) 0.9mm Al	52
4.8	Signal level in DU (digital units) as a function of square of standard deviation(noise level).	53
4.9	The measured edge spread function(ESF); (a) the noisy ESF (b) the average ESF.	55
4.10	The measured line spread function (LSF).	55
4.11	The measured modulation transfer function (MTF) in the slot and scan directions.	55
4.12	The measured noise power spectrum (NPS) for different detector exposures.	56
4.13	The detector characteristic curve used for calculation of the incremental gain.	57
4.14	The detective quantum efficiency (DQE) for different detector exposures.	58
4.15	The tissue-equivalent phantom layout specifications taken from the CIRS data sheet for Model 011A, see appendix D.	59
4.16	Phantom image from LSSM system compared to images from some existing mammography systems; (a) LSSM; (b) film-screen (GE Senographe 600T); (c) computed radiography (CR) (GE Performa); (d) digital mammography (Siemens Novation).	60
4.17	High contrast assessment using phantom images; (a) LSSM image; (b) analogue image; (c) CR image; (d) digital image; images were obtained using the technique factors shown in Table 4.3.	60
4.18	High contrast assessment using phantom images; (a) LSSM image; (b) analogue image; (c) CR image; (d) digital image.	61
4.19	Detector resolution using the Lodox Statscan line pair tool; (a) slot direction; (b) scan direction.	62
4.20	Detector additive noise for three different mammography systems.	63
4.21	The effect of detector noise on the contrast-to-noise ratio (CNR) using phantom images of a step wedge taken at an entrance exposure of $200\mu\text{Gy}$; (a) LSSM image; (b) Lodox Statscan image; (c) CNR; (d) entrance exposure.	63

4.22	The effect of increasing entrance the dose for LSSM to $1100\mu\text{Gy}$, keeping the dose for the Lodox image the same; (a)LSSM image; (b) Lodox Statscan image; (c) contrast-to-noise ratio; (d) entrance exposure.	64
5.1	The mathematical tissue-equivalent phantom for digital mammography. Areas within the phantom simulating (1) 100% glandular, 70:30 glandular/adipose, 50:50 glandular/adipose, 30:70 glandular/adipose, and 100% adipose tissues, from left to right. (2) four hemispheric regions; (3) microcalcifications of different sizes (CaCO_3); (4) line pair tools for measuring the system resolution; (5) nylon fibres of different size.	69
5.2	The effect of phantom image blur simulated for different distortion operators (Gaussian filters) on the line pair tools included in the mathematical phantom; (a), (b), and (c) are the Fourier Transforms of the applied distortion operators; (d), (e), and (f) indicate the amount of blurring for each operator.	74
5.3	A phantom image generated using 30kVp, 200mA.	76
5.4	Least squares fit of the measured pixel intensity (signal level) and the square of standard deviation (noise level).	77
5.5	The mean estimated and measured digital pixel intensity at (a) 30kVp (b) 35kVp (c) 40kVp.	79
5.6	The estimated and measured image contrast of different structures in the phantom as a function of tube voltage (a) 100% glandular tissue (b) 30:70 glandular/adipose (c) 70:30 glandular/adipose (d) 100% adipose tissue. The error bars indicate the range in the image contrast, as three image contrast readings were taken at each setting.	80
5.7	The estimated and measured image noise of 50% glandular tissue at (a) 30kVp (b) 40kVp.	81
5.8	The mathematical and the physical phantom images: (a) measured 30kVp, 80mA; (b) simulated 30kVp, 80mA; (c) measured 35kVp, 100mA; (d) simulated 35kVp, 100mA; (e) measured 40kVp, 64mA; (f) simulated 40kVp, 64mA. The nylon fibres in the mathematical phantom were simulated having the same diameters as those in the physical phantom, but in a vertical rather than diagonal direction for ease of implementation.	82
5.9	Measured and simulated contrast-to-noise ratio (CNR) for different tissues within the phantom image at 30kVp tube voltage; (a) 100% glandular tissue; (b) 70% glandular tissue; (c) 30% glandular tissue; (d) 100% adipose tissue. The error bars indicate the range in CNR in each region of interest.	83
5.10	Measured and simulated contrast-to-noise ratio (CNR) for different tissues in the phantom image at 40 kVp tube voltage (a) 100% glandular tissue; (b) 70% glandular tissue; (c) 100% adipose tissue. The error bars indicate the range in CNR in each region of interest.	84

5.11	Differences between the measured and simulated CNR for different kVps as a percentage of their mean (a) 30 kVp; and (b) 40 kVp	85
6.1	Image simulated using different optical coupling. Image of (a) step wedge using direct coupling; (b) step wedge using fibre optic taper; (c) microcalcifications and tumour masses using direct coupling; (d) microcalcifications and tumour masses using fibre optic taper. The bar plot indicates the improvement of output signal in terms of CNR and DQE at the same detector exposure, of direct coupling compared to indirect coupling. . . .	90
6.2	The effect of reducing the additive noise on detector performance. Simulated images using additive noise of (a) 1000 e- (b) 4000 e- . The bar plot indicates the improvement of detector performance in terms of CNR by reducing the additive noise (4000 e-).	91
6.3	Bragg's law , x-ray diffraction	93
6.4	The estimated MXF monochromatic beams at different tuned energy levels and the unfiltered spectrum.	94
6.5	The effect of monochromatic beams at different tuned energy levels on the image contrast and the system DQE. The image quality parameters were calculated using the simulation model (<i>MammoModel</i> described in Appendix A).	94
6.6	The image performance of the polychromatic (left) and the monochromatic (right) beams at tube voltage 50kVp and tube current 100mA - the polychromatic beam is the unfiltered spectrum shown in Figure 6.4 and the monochromatic, is a beam tuned to 19 keV; (a) and (b) the performance using the same technique factors; (c) and (d) doubling the scanning time of the monochromatic beam; (e) and (f) tripling the scanning time of the monochromatic beam; (g) and (h) doubling the scanning time and using FWHM of 25% rather than 10% for the monochromatic beam.	97
7.1	Small object embedded on a uniform material	100
7.2	Effect of detection energy threshold on the DQE at zero spatial frequency for W:Al(0.5mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.	104
7.3	Effect of detection energy threshold on the DQE at zero spatial frequency for Mo:Mo(0.03mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.	105
7.4	Effect of detection energy threshold on the FOM for W:Al(0.5mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.	106

7.5	Effect of detection energy threshold on the FOM for Mo:Mo(0.03mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm. . .	107
7.6	Effect of detection energy threshold on the FOM for tungsten spectra at various x-ray tube peak voltages for a 50% glandular 8cm compressed breast, the spectrum filtered with (a) Al (0.5mm), (b) Mo (0.03mm), (c) Rh (0.03mm).	108
A.1	The main graphical user interface (GUI) program.	115
A.2	Slot scanning mammography system.	117
A.3	Transmission efficiency of the fiber optic taper FOT.	121
A.4	Window for the calculation of the detector additive noise.	122
A.5	MammoModel Help window.	124
A.6	Multispectrum optimization.	125
A.7	PhantomUser Help window.	128
A.8	Photon counting detector optimization.	129
A.9	The OptimumThreshold Help window.	131
B.1	(a) beam shaping collimator; (b) mounted frame	132
B.2	(a) beam shaping collimator; (b) slot collimator	133
B.3	(a) LSSM detector; (b) Three CCD cameras; (b) CCD acquisition and control card	133

List of Tables

3.1	The technical specifications of the Hamamatsu CCD (CCDS8658, 2004) .	30
3.2	The estimated charge well capacity and the dynamic range for the Hamamatsu detector; technical data are taken from (CCDS8658, 2004).	31
3.3	Slot scanning mammography specifications	32
4.1	Summary of LSSM specifications; components are shown in Figure 4.2. .	47
4.2	The half value layer for different tube voltages.	52
4.3	Summary of the parameters used for digital images obtained of the physical phantom; the figures in brackets represent tube current mA rather than exposure time mAs. The minimum CaCO_3 grain and hemispherical mass sizes visible at these technique factors are shown.	65
5.1	Mathematical phantom specifications	70
5.2	Parameters used in obtaining digital images of the physical phantom. . .	77
6.1	Summary of LSSM system specifications, SDD is source-to-detector distance	89
A.1	The coefficients of the fitted equations used for calculation of SPR for slot scanning mammography	123

Chapter 1

Introduction

1.1 Project purpose and scope

Mammography plays an important role in the detection of tumours and microcalcifications in the breast and is used in screening for early detection of breast cancer (Kopans *et al.*, 1984; Nishikawa *et al.*, 1987; Andrew *et al.*, 1993; Fahrig and Yaffe, 1994; Smith *et al.*, 2004). Early treatment can increase the survival rate of breast cancer patients (Nathalie, 2007). Screen-film mammography has been used as a breast screening system since 1960 (Egan, 1960). Although advances in screen-film mammography and film processing techniques have improved the image quality so that film-based systems provide excellent spatial resolution (>10 lp/mm), they are limited in their dynamic range and contrast (Nishikawa *et al.*, 1987). Digital mammography has been under investigation for over a decade to overcome the limitations of traditional screen-film mammography. The advantages of digital radiography over screen-film radiography are those associated with dose reduction and image processing in terms of enhancing regions of interest, in addition to the capability of image storage and retrieval. In January 2000 Senographe 2000D, the first digital mammography system was approved by the US Food and Drug Administration (FDA) to be used for routine breast cancer screening instead of the standard screen-film mammography system.

Digital systems using a narrow fan beam scanned over the patient in synchrony with a slot detector provide excellent scatter rejection. These systems eliminate the need for a grid and provide high image resolution and contrast with low radiation dose to the

patient. They therefore have the potential to be used in mammography systems, for dose reduction and improved image quality. Linear slit or slot scanning radiography using a narrow fan beam has been reported to show a significantly lower dose compared to that of conventional x-ray systems (deVilliers and deJager, 2003; Irving *et al.*, 2008). Slit scanning systems have one-pixel wide detectors, while slot scanning detectors are wider (Lundqvist, 2003; Smith *et al.*, 2004).

Computer simulation models are considered important tools for the design and optimization of digital x-ray systems. Several computer models have been developed to aid in the design and in image performance evaluation of digital systems (Fahrig and Yaffe, 1994; Siewerdsen *et al.*, 2004). These models depend on energy transport and cascaded linear analysis of the propagation of signal and noise through the imaging system. The models predict the signal to noise ratio (SNR) and detective quantum efficiency (DQE) for a given organ dose (the breast is of interest in this thesis), and therefore allow detector optimization for different spectra and different imaging tasks.

The assessment of system performance using test phantoms also plays an important role in the design and optimization of digital mammography. Several physical test phantoms have been developed for performance assessment of digital mammography, although these have limitations in their application, and a need exists for digital test phantoms.

Photon counting detectors in digital radiography have a number of advantages over charge integrated detectors. These advantages are: the improvement of image quality, the dose efficient rejection of scattered radiation with the implementation of energy discrimination in the sensor and the effective rejection of electronic noise and Swank noise (Tapiovaara and Wagner, 1985; Gobbi *et al.*, 1999; Lundqvist, 2003; Manuilskiy *et al.*, 2004). Photon counting detectors therefore have a role to play in digital mammography and have been tested for this application (Aslund *et al.*, 2007).

This PhD project formed part of an NIH-funded project (5R21CA101705-02) “Low dose digital X-rays for mammographic screening” conducted in the MRC/UCT Medical Imaging Research Unit, University of Cape Town. The project described in this report had the following objectives:

- To develop computer modelling and optimization tools for system design and performance analysis in digital mammography.

- To use the tools for the design of a mammography system with low radiation exposure.
- To use the tools to improve image quality with respect to existing mammography systems in terms of modulation transfer function and detective quantum efficiency.

This document reports on the design, evaluation, and optimization of a prototype linear slot scanning mammography system. Computational models were developed for system design and optimization. The system requirements were estimated using a computer model titled *MammoModel* similar to that described by Siewerdsen *et al.* (2004), but adapted for linear slot scanning digital mammography. A computer model titled *PhantomUser* was developed for optimization of the LSSM detector and for spectrum optimization using a mathematical breast phantom. Another computer model titled *OptimumThreshold* was developed for the selection of optimum detection thresholds and examining the effect of the anode-filter combination on the performance of a photon counting detector.

1.2 Document overview

This document presents computer simulation tools for the design and optimization of a linear slot scanning mammography (LSSM) system. Because the proposed LSSM has the same configuration as the Lodox Statscan (Lodox System, Sandton, South Africa), the Lodox Statscan was used together with the LSSM system for validation of the models and for detector evaluation. Knowledge of the principles of x-ray imaging is assumed throughout the document.

Chapter 2 gives an overview of breast anatomy, breast screening, breast cancer, and the modalities available for imaging the breast. The existing x-ray imaging analysis methods used in the design of our LSSM system are also described in this chapter.

Chapter 3 presents the estimated detector performance and describes system design specifications and requirements for a prototype LSSM system. The system was designed to provide a linear scanning geometry with a narrow fan beam to offer good scatter rejection.

Chapter 4 presents the description and evaluation of the prototype LSSM system built in the MRC/UCT Medical Imaging Research Unit based on the the system design and requirements described in Chapter 3. A commercially available tissue-equivalent phantom designed for mammography was used for system evaluation.

Chapter 5 introduces a tissue-equivalent mathematical phantom for digital mammography developed for optimization of the LSSM system. The model was validated using the Lodox Statscan and the LSSM system.

Chapter 6 presents the application of the mathematical phantom developed in the previous chapter. Different detector configurations and the effects of electronic noise on system performance were investigated. Spectrum optimization using a monochromatic x-ray beam was also examined.

Chapter 7 presents a computer model developed to assist in the implementation of novel algorithms developed by Marchal *et al.* (2006) for the selection of detection thresholds in single-photon counting in digital mammography. The image quality parameters detective quantum efficiency and figure of merit were used to examine the effect of different detection thresholds and different anode-filter combinations on the performance of digital mammography using different imaging tasks and different x-ray spectra.

Chapter 8 summarises the preceding chapters and discusses the relevance of the work presented. Possible directions for future work are explored.

Some of the work presented appears in the following journal articles and conference proceedings:

- K. Hussein, C. L. Vaughan and T. S. Douglas, 2009. Modeling, validation and application of a mathematical tissue-equivalent breast phantom for linear slot-scanning digital mammography. *Physics in Medicine and Biology* 54: 1533-1553 (currently available in electronic form).
- K. Hussein, J. Marchal, B.S. Spottiswoode, S. Steiner, T. S. Douglas and C. L. Vaughan, 2007. Theoretical evaluation of linear slot scanning digital mammography. In: *Proc. 6th Edward Bouchet International Conference on Physics and Technology for Sustainable Development in Africa (EBASI)*, iThemba LABS, Cape Town, South Africa, 24 - 26th of January.

- J. P. Marchal, K. Hussein, L. R. John and C. L. Vaughan, 2006. Photon counting threshold optimization in mammography. *Nuclear Instruments and Methods in Physics Research*, A 563, 137-141.

University of Cape Town

Chapter 2

Background Theory

2.1 Anatomy of the breast

The breast is composed of 30-40% glandular tissue and the rest is adipose tissue. The breast is surrounded by an outer layer of superficial skin (0.5 to 2.0 mm thick)(Geise and Palchevsky, 1996). The glandular tissue is the most sensitive tissue in the breast to radiation. Glandular tissue consists of a number of ducts, each of which defines a segment or lobe of the gland. Between 8 and 20 ducts become superficial at the surface of the nipple tissue. The density of the breast depends on the degree of glandular tissue present in the breast and this changes throughout a woman's lifetime. The anatomy of the breast is depicted in Figure 2.1.

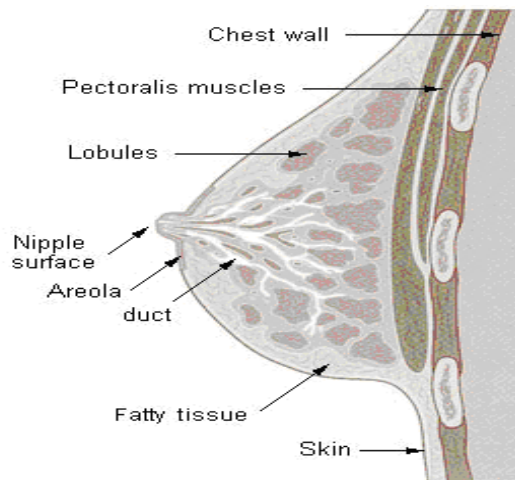


Figure 2.1: Breast anatomy, adapted from (BreastCancer, 2007)

2.2 Screening for breast cancer

Many factors influence the incidence of breast cancer such as genetic factors, exposure to carcinogens and overexposure to radiation (NIEHS, 2008). Mammography is the most common procedure for the screening and diagnosis of breast cancer (Smith *et al.*, 2004), and is considered a useful tool in early detection and treatment. During mammography, the breast is compressed to spread the breast tissue so that the maximum amount of tissue can be imaged and at minimum radiation; this may result in patient discomfort and pain. The average compressed breast thickness is in the range between 4 and 5cm.

2.2.1 Screen-film mammography

Screen-film mammography has been used for breast cancer screening since it was described in 1960 (Egan, 1960). Although advances in screen-film mammography and film processing techniques have improved the image quality so that screen-film mammography systems provide excellent spatial resolution of more than 10 lp/mm, they are limited in their dynamic range (have a non-linear response to the photon flux) and contrast. Therefore it is difficult to see all the different structures in the breast such as dense glandular tissue and adipose tissue at the same exposure. Other disadvantages of screen-film mammography are film granularity, which affects the detective quantum efficiency and the visibility of microcalcifications, and the difficulties of chemical processing (Destounis, 2008).

Molybdenum-molybdenum is the most common anode-filter combination in screen-film mammography. The screens used in screen-film mammography are made of rare earth phosphors, e.g. Gadolinium Oxysulphide ($\text{Gd}_2\text{O}_2\text{S:Tb}$) (Kopans, 1986; Haus, 1987), which produce light photons when exposed to x-ray energy. The film captures the photon light to produce analogue images.

2.2.2 Digital mammography

Digital mammography has been under investigation for over a decade in an attempt to overcome the limitations of traditional screen-film mammography. The advantages of digital radiography over screen-film radiography are those associated with dose reduc-

tion, image processing in terms of enhancing regions of interest, and the capability of image storage and retrieval. In January 2000 Senographe 2000D, the first digital mammography system was approved by US Food and Drug Administration (FDA) to be used for routine breast cancer screening (NCI, 2004). There are several full-field digital radiography systems reported or under development to replace the conventional screen-film image detector. These systems are based on amorphous silicon (a-Si) or amorphous selenium (a-Se) flat panels or charge-coupled devices (CCDs) (Rowlands and Hunter, 1991; Stierstorfer and Spahn, 1999). The incident x-rays interact with the scintillator, and the x-ray interactions result in the deposition of energy from which light photons are generated. A fraction of these light photons exit from the bottom surface of the scintillator and are transmitted through the optical coupling media (fibre optic taper and fibre optic plate), collected by CCDs and converted into electron-holes which contribute to the measured signal. The detector components of a digital system can be seen in Figure 2.2. Digital detectors can be classified into different categories according to the way they pick up the incident x-ray photons and convert them into an electrical signal. These types of detectors are summarised in the following sections.

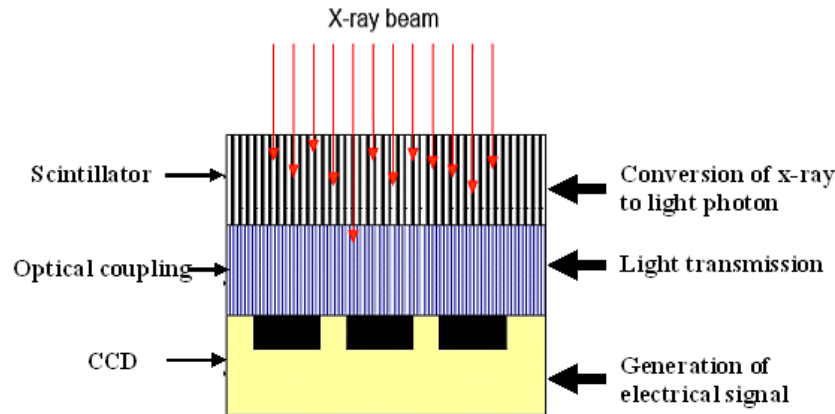


Figure 2.2: The detector components of a digital system (on the left) and their function (on the right).

2.2.2.1 Indirect detectors

In indirect detectors, known as flat panel detectors, the x-ray photons are converted into electrical signals in two steps; firstly the x-rays are converted to light photons using a scintillator such as cesium iodide doped with thallium (CsI:TI) or a rare earth phosphor,

e.g. gadolinium oxysulphide ($\text{Gd}_2\text{O}_2\text{S:Tb}$) (Jing *et al.*, 1996). Light photons are then converted into electrical signals by using amorphous silicon thin-film transistors (TFT), photodiode arrays or imaging devices such as CCDs. The indirect detector performance is a trade-off between interaction efficiency, and spatial resolution. The thicker the scintillator, the higher the quantum efficiency and the lower the spatial resolution, and vice versa. The scintillator is optically directly coupled to the TFT/photodiode array or CCD to increase the transmission efficiency of the detector, or optically indirectly coupled using a fibre optic taper to reduce the cost of the detector but also the transmission efficiency of the detector. An example of an indirect detector is the GE Senographe 2000D approved in January 2000.

2.2.2.2 Direct detectors

In direct detectors, the x-ray photons are converted to electrical signals in one step. A photoconductor absorbs the x-rays and directly converts them into electron-hole pairs. The charge generated in a photoconductor is collected in the capacitor of the pixel using an applied electric field. There is no lateral charge spreading, which results in better spatial resolution than in indirect detectors even for thicker photoconductors. There is no trade-off between the resolution and absorption efficiency. The photoconductor used in direct conversion detectors is amorphous selenium, which has high quantum efficiency in the mammography range. An example of a direct conversion detector is the Lorad/Hologic Selenia approved by the FDA in October 2002.

2.2.2.3 Scanning radiographic detectors

Digital systems that use a narrow fan beam with a post-patient slit/slot detector provide excellent scatter rejection. The detector of slit scanners is only 1 pixel wide, while the slot scanners have detectors up to 1cm wide (Lundqvist, 2003; Smith *et al.*, 2004). Two configurations are used in the scanning radiographic systems, namely, linear slit/slot scanning radiography (LSSR) in which the x-ray source moves linearly, synchronized with the detector; and fixed source slit/slot scanning radiography (FSSR), in which a fixed source with a collimated fan beam sweeping a radial path is used to scan the patient. These systems eliminate the need for a grid, which is normally used in conventional mammography systems to reduce detection of scattered radiation, and provide

high image resolution and contrast with a low radiation dose to the patient. The slit/slot detector consists of a Thallium-activated Cesium Iodide (CsI:TI) or rare earth phosphor, e.g. Gadolinium Oxysulphide ($\text{Gd}_2\text{O}_2\text{S:Tb}$), scintillator coupled to CCDs operating in time delay integration (TDI) mode (explained in Section 3.2.4) to provide a good balance between high absorption efficiency and high spatial resolution. The scintillator is coupled to the CCD through a fibre-optic plate or taper in order to improve the efficiency and reduce the effective size of the detector. In radiography, it has been reported that LSSR has a significantly lower dose compared to that of conventional x-ray systems (deVilliers and deJager, 2003; Irving *et al.*, 2008). In mammography, the Fischer system (SenoScan) (Tesic *et al.*, 1999) that uses the FSSR configuration showed a better resolution and detection efficiency but with a higher dose compared to other digital mammography systems.

2.3 Image quality parameters

Measurement of the image quality produced by medical imaging systems is mainly based on statistical detection theory assuming an ideal observer (ICRU, 1994). There are several image quality parameters used for quantifying the performance of x-ray imaging systems such as signal-to-noise ratio (SNR), noise equivalent quanta (NEQ) and detective quantum efficiency (DQE). For example, if a small detail needs to be imaged, then imaging system performance at high spatial frequencies may be a more important factor than imaging system performance at low spatial frequencies, and vice versa if the visibility of large, low-contrast objects is required (ICRU, 1994).

2.3.1 Signal-to-noise ratio (SNR)

The signal-to-noise ratio measures the detectability of specific details in the image. Rose (1973) provided an early definition of a contrast-to-noise ratio (CNR): the ratio of object contrast against a uniform background to the standard deviation of the image noise. Figure 2.3 illustrates the Rose model. The symbols are described below. According to Rose (1973), the contrast C of the target tissue over an area A against a uniform background B of the same area can be written as the ratio of the object signal and

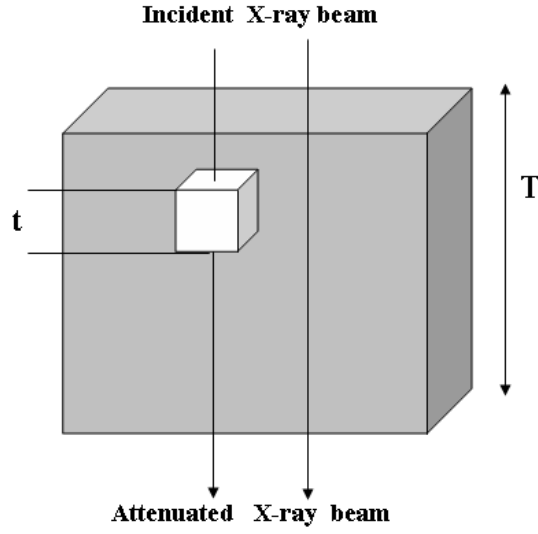


Figure 2.3: The Rose model

background;

$$C = \frac{Q_B - Q_A}{Q_B} \quad (2.1)$$

where Q_A and Q_B are the attenuated photon intensities through the target and outside the target respectively, and they can be defined as follows:

$$Q_B = \bar{\Psi}_E \varepsilon e^{-\mu_B T} \quad (2.2)$$

$$Q_A = \bar{\Psi}_E \varepsilon e^{-\mu_B (T-t) - \mu_A t} \quad (2.3)$$

where $\bar{\Psi}_E$ is the incident photon fluence; ε is detector quantum efficiency; μ_A and μ_B are the energy-dependent linear attenuation coefficients of the target and background area respectively; and T is breast thickness. The contrast can be written as:

$$C = [1 - e^{-(\mu_A - \mu_B)t}] \quad (2.4)$$

The contrast-to-noise ratio is given by:

$$CNR = \frac{A(Q_B - Q_A)}{\sqrt{AQ_B}} = C\sqrt{AQ_B} = C\sqrt{A\bar{\Psi}_E \varepsilon e^{-\mu_B T}} = C\sqrt{A\bar{\Psi}_0 \varepsilon} \quad (2.5)$$

where $\bar{\Psi}_0$ is the mean incident fluence on the detector.

Rose (1973) assumes a statistically uncorrelated distribution of image quanta in his model; in reality the contribution of the scattered x-rays and secondary quanta in the detection system may introduce statistical correlation in the image quanta. The more general definition of signal-to-noise ratio for an ideal observer, which accounts for signal and noise propagation in a linear system and the frequency-dependent task of the detected signal, as described in ICRU (1994), is:

$$SNR_I^2 = \int_0^{f_{Nyq}} |\Delta \tilde{f}(f)|^2 NEQ_S(f) df \quad (2.6)$$

The frequency-dependent task $\Delta \tilde{f}(v)$ defines the detectability of the object against a uniform background; $NEQ_S(f)$ is the noise equivalent quanta of the imaging system and describes the effective number of incident x-ray quanta contributing to the signal-to-noise ratio (SNR) of the output image. The fraction of incident quanta that contributes to SNR is characterized by the system detective quantum efficiency ($DQE_S(f)$), which is related to the noise equivalent quanta (NEQ), as described in ICRU (1994), by:

$$NEQ_S(f) = \frac{G^2 MTF_S^2(f)}{NPS_S(f)} = DQE(f) \times \tilde{\Psi}_0 \quad (2.7)$$

This allows equation 2.6 to be written as

$$SNR_I^2 = \int_0^{f_{Nyq}} |\Delta \tilde{f}(f)|^2 \frac{G^2 MTF_S^2(f)}{NPS_S(f)} df \quad (2.8)$$

$MTF_S(f)$ is the system modulation transfer function; $NPS_S(f)$ is the total noise power spectrum; G is the large area gain representing the detector signal. The Nyquist frequency is given by Siewerdsen and Jaffray (2000)

$$f_{Nyq} = \frac{1}{2 \times \text{lesionsize}} \quad (2.9)$$

The modulation transfer function $MTF_S(f)$ and the noise power spectrum $NPS_S(f)$ of the system can be estimated using the linear cascaded model as discussed in the following section.

2.3.2 Linear Cascaded Model

The propagation of signal and noise throughout the entire x-ray imaging system to analyze the detector performance has been studied using a cascaded serial model (Rabbani *et al.*, 1987; Rabbani and Van Metter, 1989; Cunningham, 1994; Siewerdsen and Antonuk, 1998; Siewerdsen *et al.*, 1998; Stierstorfer and Spahn, 1999; Siewerdsen and Jaffray, 2000; Tkaczyk *et al.*, 2001; Cunningham *et al.*, 2002; Neitzel *et al.*, 2004). The transfer of signal and noise can predict the overall system $MTF_S(f)$ and $NPS_S(f)$; this requires the system to have a linear and shift-invariant signal response and stationary noise process, which are expressed in terms of noise power spectrum. The propagation of signal and noise through the entire system can be divided into different stages as illustrated in Figure 2.4. Each stage in the image chain represents a process involved in the formation of the final image.

These stages represent a gain stage, a spreading stage, or blurring associated with detector element size. Stage 0 represents the incidence of quanta on the CsI scintillator and interaction of x-rays with the CsI scintillator with an interaction efficiency of $\bar{\eta}_a$. The interaction results in the absorption of x-rays and the generation of light photons. Since the generation of light and its associated noise is a random process, the next stage represents a stochastic amplification process where the conversion gain g_{scr}^- and the associated noise in terms of gain-variance g_{scr}^{-2} characterize the transfer of signal and noise in this stage. The stochastic blurring that occurs in this stage before the light photons leave the exit windows of the scintillator is represented by the modulation transfer function of the scintillator $MTF_{scr}(f)$. The next stage represents the effect of transmission and reflection of a light photon before it reaches the CCD and the generation efficiency of signal charges (e-h pair) at the CCD, namely the total coupling efficiency of fibre optics with the CCD $\bar{\eta}_C$. The deterministic blurring caused by the finite pixel size of the CCD in this stage is characterized by the presampling modulation transfer function MTF_{samp} . Finally, the additive electronic noise, which is dependent on the amplifier chosen, the external circuit impedance, the gain, the circuit bandwidth, and the ambient temperature, is added to the final output signal and noise.

Using the results of Rabbani *et al.* (1987) the noise power spectrum in the i th stage, having gain only (no spreading), is given by (Cunningham, 1994):

$$NPS_i = \bar{g}_i^2 NPS_{i-1} + \sigma_{g_i}^2 Q_{i-1} \quad (2.10)$$

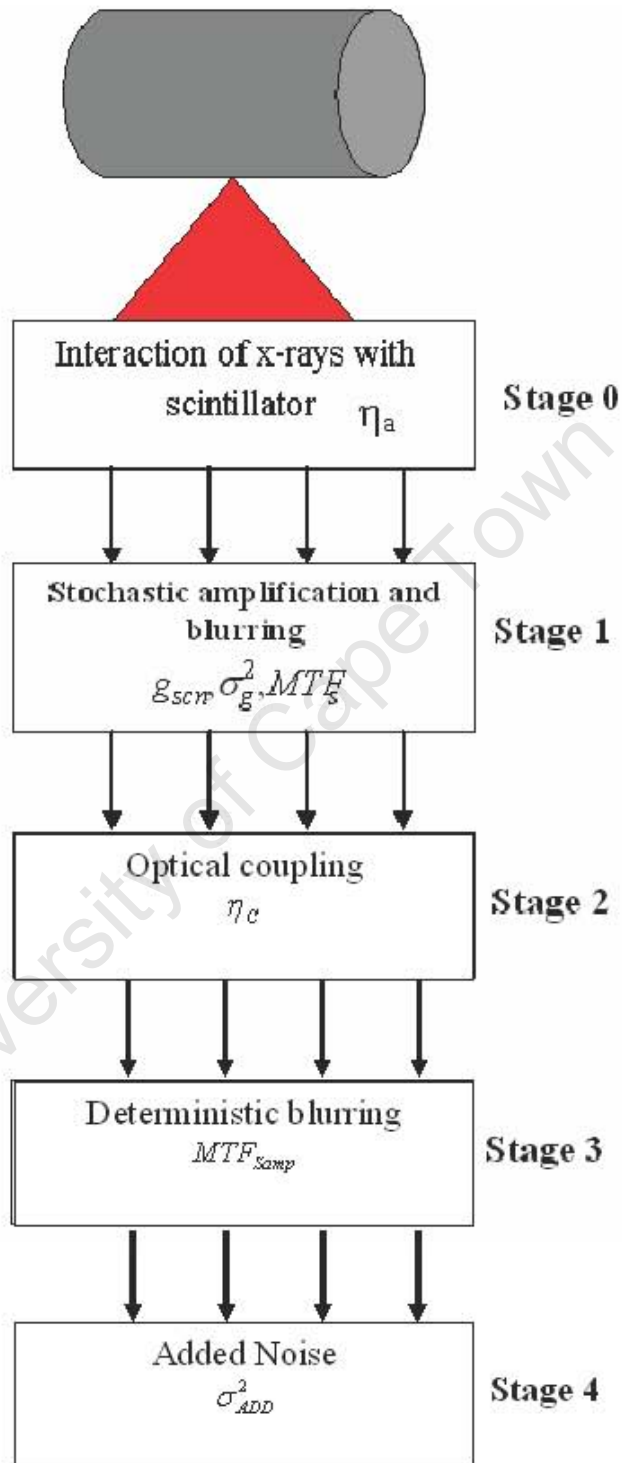


Figure 2.4: Linear cascaded model

Q_i is the mean number of quanta at the i th stage; $\bar{g}_i$ is stage gain; $\sigma_{g_i}^2$ is the variance of the gain, which can also be described by a Poisson excess ε . The Poisson excess is the amount by which the gain fluctuation exceeds a Poisson distributed gain fluctuation. In a Poisson distributed gain fluctuation, the variance is equal to the mean number of quanta. The Poisson excess is given by the following expression:

$$\varepsilon = \frac{\sigma_{g_i}^2}{\bar{g}_i} - 1 \quad (2.11)$$

The noise power spectrum in each stage can be restated as follows:

$$NPS_i = \bar{g}_i^2 NPS_{i-1} + Q_{i-1} \bar{g}_i (\varepsilon + 1) \quad (2.12)$$

In the quantum scatter stage the noise power spectrum is modulated by the squared modulation transfer function, which is the Fourier transform of the stage point spread function (PSF). Starting from the expression described by (Rabbani *et al.*, 1987) and including the effect of quantum scatter, the noise power spectrum of the scintillator NPS_{scn} can be restated as follows:

$$NPS_{scr}(f) = \bar{g}_{scr}^2 \bar{\eta}_a \bar{\Psi}_0 \left(1 + \frac{\varepsilon}{\bar{g}_{scr}}\right) MTF_{scr}^2(f) + \bar{g}_{scr} \bar{\eta}_a \bar{\Psi}_0 \quad (2.13)$$

$\bar{g}_{scr}$ is the number of light photons emitted by the screen per absorbed x-ray photon; MTF_{scr} is the modulation transfer function of the screen; and ε is the noise in excess of Poisson noise associated with statistical variation in the conversion of x-rays into light quanta.

Introducing the effect of optical coupling efficiency, which is characterized by the coupling efficiency of the fibre optics, results in presampling $NPS_{presamp}$.

$$\begin{aligned} NPS_{presamp}(f) &= \bar{\eta}_C^2 NPS_{scr}(f) + \bar{g}_{scr} \bar{\eta}_a \bar{\eta}_C \bar{\Psi}_0 (1 - \bar{\eta}_C) \\ &= \bar{\eta}_C^2 \bar{g}_{scr}^2 \bar{\eta}_a \bar{\Psi}_0 \left(1 + \frac{\varepsilon}{\bar{g}_{scr}}\right) MTF_{scr}^2(f) + \bar{\eta}_C^2 \bar{g}_{scr} \bar{\eta}_a \bar{\Psi}_0 + \bar{g}_{scr} \bar{\eta}_a \bar{\eta}_C \bar{\Psi}_0 (1 - \bar{\eta}_C) \\ &= \bar{\eta}_C \bar{g}_{scr} \bar{\eta}_a \bar{\Psi}_0 [\bar{\eta}_C (\bar{g}_{scr} + \varepsilon) MTF_{scr}^2(f) + 1] \end{aligned} \quad (2.14)$$

The effect of blurring due to integration of x-ray quanta by the CCD pixel aperture, which is characterized by MTF_{samp} and the pixel size a_{pix} , can be added as defined by Siewerdsen and Jaffray (2000):

$$NPS_{samp}(f) = a_{pix}^4 NPS_{presamp}(f) MTF_{samp}^2(f)$$

$$NPS_{samp}(f) = a_{pix}^4 \bar{\eta}_C \bar{g}_{scr} \bar{\eta}_a \bar{\Psi}_0 [\bar{\eta}_C (g_{scr}^- + \varepsilon) MTF_{scr}^2(f) + 1] MTF_{samp}^2(f) \quad (2.15)$$

The $NPS_{samp}(f)$ equation can be modified to include the anti-scatter grid, the slot scanning geometry and the additive electronic noise, which gives the total noise power spectrum, $NPS_T(f)$, slow scintillation decaying component(afterglow)

$$NPS_T(f) = [T_P + T_S \times SPR] [a_{pix}^4 \bar{\eta}_C \bar{g}_{scr} \bar{\eta}_a \bar{\Psi}_0 [\bar{\eta}_C (g_{scr}^- + \varepsilon) MTF_{scr}^2(f) + 1] MTF_{samp}^2(f) + \sigma_{add}^2] \quad (2.16)$$

where T_S and T_P are the scatter and primary transmission factors that characterize the anti-scatter and slot scanning geometry, for a slot scanning system without a grid, T_S and T_P are equal to 1; SPR is the scatter-to-primary ratio at the detector. The afterglow (slow scintillation decaying component) and MTF degradation due to the temporal response of the scintillator is neglected due to the fast response of the CsI scintillator.

2.3.2.1 System gain

The total detector gain is the product of the individual gains at the different stages. The individual gains that contribute to the total detector gain consist of the absorption efficiency of the scintillator, the detector conversion g_{scr}^- , and the total coupling efficiency $\bar{\eta}_C$. Several components affect the total coupling efficiency, namely the transmission efficiency of the fibre optic taper (η_{FOT}), the transmission efficiency of the fibre optic plate (η_{FOP}), the coupling efficiency of the fibre optic taper to the fibre optic plate ($\eta_{FOTtoFOP}$), and the charge coupled device quantum efficiency (η_{CCD}). The different gain components can be estimated from the following equations:

$$\bar{\eta}_a = 1 - \exp[-\mu_{Det}x] \quad (2.17)$$

where μ_{Det} is the total linear attenuation of the scintillator; x is the scintillator thickness;

$$g_{scr}^- = \frac{E}{W} \quad (2.18)$$

where E is the energy of the absorbed photon; W is the average energy required to generate a light photon (Zhao *et al.*, 2004; Rowlands and Yorkston, 2000);

$$\bar{\eta}_C = \eta_{FOT} \times \eta_{FOTtoFOP} \times \eta_{CCD} \times \eta_{FOP} \quad (2.19)$$

The total gain of the detector G in electrons/photon/mm² pixel is then given by:

$$G = a_{pix}^2 \times \bar{\eta}_C \times g_{scr}^- \times \bar{\eta}_a \quad (2.20)$$

2.3.2.2 System MTF

The detector spatial resolution, which is determined by the modulation transfer function (MTF) in the scan and slot direction, can be estimated from the total blurring occurring in the linear cascaded stages.

System MTF in the slot direction

The system resolution in the slot direction is limited by geometrical blurring due to the finite focal spot size in the slot direction MTF_{Focal} , which can be estimated from the intensity distribution of the focal spot (discussed in detail in the following chapter), the integration of quanta over the detector element size in the slot direction $MTF_{samp_x}(f)$, and the spatial resolution of the screen $MTF_{scr}(f)$. The total modulation transfer function MTF_{Slot} is the product of all the MTF contributions in the slot direction.

$$MTF_{Slot}(f) = MTF_{Focal}(f) \times MTF_{samp}(f) \times MTF_{scr}(f) \quad (2.21)$$

The $MTF_{scr}(f)$ can be estimated using the following fitting equation (Ganguly *et al.*, 2003).

$$MTF_{scr}(f) = \frac{\pi}{4} \left[a \{ e^{-bDf} - \tan^{-1}(e^{-bDf}) \} + y_0 \{ 2 - \frac{\pi}{4} \} \right] \frac{1}{100} \quad (2.22)$$

where $a = 99.99$, $b = 1.528$, $y_0 = 0.9087$, and D is the thickness of the scintillator (Ganguly *et al.*, 2003).

The presampling MTF_{samp} which determines the blurring due to the integration of quanta in the CCD pixel is given by the sinc function (Siewerdsen and Antonuk, 1998)

$$MTF_{samp_x}(f) = \text{sinc}(a_{pd_x}f) \quad (2.23)$$

$$MTF_{samp_y}(f) = \text{sinc}(a_{pd_y}f) \quad (2.24)$$

a_{pd_x} and a_{pd_y} are the effective pixel size in the slot and scan directions respectively. The effective pixel size is characterized by the fill factor (*ffactor*) as follows:

$$a_{pd} = a_{pixel} \times \text{ffactor} \quad (2.25)$$

System MTF in the scan direction

In the scan direction, the detector spatial resolution is again determined by the geometrical blurring, the spatial resolution of the screen, and the integration of quanta by the detector element size in the scan direction $MTF_{samp_y}(f)$. Other factors add to the degradation of the total MTF in the scan direction, namely: the blurring due to the scintillator decay characteristics MTF_{Decay} and the asynchronism between the scanning velocity and the charge transfer rate $MTF_{\Delta v}$. Due to the small slot width (6mm) and long depth (7mm) of the post-collimator of the proposed mammography system, the effect of geometrical blurring and penumbra were neglected in the estimation of system MTF in the scan direction. The total modulation transfer function MTF_{Scan} is the product of all the MTF contributions in the scan direction.

$$MTF_{Scan}(f) = MTF_{Decay}(f) \times MTF_{scr}(f) \times MTF_{samp_y}(f) \times MTF_{\Delta v}(f) \quad (2.26)$$

The blurring due to the scintillator decay characteristic (MTF_{Decay}) is given by (Mainprize and Yaffe, 1996)

$$MTF_{Decay}(f) = \frac{1}{\sqrt{1 + (\frac{2\pi f}{k})^2}} \quad (2.27)$$

The resolution degradation due to the non-synchronism between the scanning velocity and the charge transfer rate $MTF_{\Delta v}$ is given by Barbe (1976):

$$MTF_{\Delta v}(f) = \text{sinc} \left(N \Delta y \frac{\Delta v}{v_d} f \right) \quad (2.28)$$

N is number of pixels in the scan direction, Δy is pixel size in the scan direction, Δv is the difference between the scan speed and the charge transfer speed, and f is the spatial frequency.

2.3.2.3 Task function

The task function $\Delta \tilde{f}(f)$ is defined as the square of Fourier transform component of the difference of the object and the background ICRU (1994):

$$f_{task} = |\vartheta(object) - \vartheta(Background)|^2 \quad (2.29)$$

The object can be represented by a Gaussian distribution characterized by the size of the object. For a large low-contrast object the difference can be written in terms of contrast C over an area A_{obj} against the background noise.

$$\Delta \tilde{f}(f) \cong C \tilde{f}(f) \quad (2.30)$$

The contrast C is given by:

$$C = 1 - e^{-\Delta \mu t} \quad (2.31)$$

where $\Delta \mu$ is the difference between the linear attenuation of the target tissue and the background and t is the thickness of the target tissue. An object of area a_{obj} against a uniform background is given a Gaussian distribution function $f(x)$ in the spatial domain and $f(f)$ in frequency domain :

$$f(x) = e^{-\pi x^2} a_{obj}^2$$

$$\Rightarrow$$

$$f(f) = a_{obj} e^{-a_{obj} \pi f^2}$$

Then $\Delta \tilde{f}(f)$ can be written as follows:

$$\Delta \tilde{f}(f) \approx C \times a_{obj} e^{-a_{obj} \pi f^2} \quad (2.32)$$

2.3.3 SNR and DQE

Substituting Equations 2.16, 2.21, 2.20, and 2.32 in Equation 2.8, the signal-to-noise ratio can be written as follows:

$$SNR_I^2 = \int_0^{f_{Nyq}} |\Delta \tilde{f}(f)|^2 \frac{G^2 MTF_S^2(f)}{NPS_S(f)} df$$

$$= \int_0^{f_{Nyq}} |a_{obj} C e^{-a\pi f^2}|^2 \left\{ \frac{T_P \times (a_{pix}^2 \times \bar{\eta}_C \times g_{scn} \times \bar{\eta}_a)^2 MTF_{FScan}^2(f) MTF_{samp}^2(f) MTF_{\Delta v}^2(f) MTF_{\Delta v}(f)^2}{a_{pix}^4 \bar{\eta}_C g_{scn} \bar{\eta}_a \bar{\Psi}_0 [1 + \frac{T_S}{T_P} SPR] [\bar{\eta}_C (g_{scn} + \varepsilon) MTF_{scn}^2(f) + 1] MTF_{samp}^2(f) + \sigma_{add}^2} \right\} df \quad (2.33)$$

a_{obj}^2 represents the area of the object A_{obj} being imaged. The second term is, in practice, equivalent to the noise equivalent quanta ($NEQ_S(f)$) which is equal to:

$$NEQ_S(f) = DQE(f) \times \tilde{\Psi}_0 \quad (2.34)$$

Equation 2.33 can also be written as:

$$SNR_I = C \sqrt{\tilde{\Psi}_0 A_{obj} \int_0^{f_{Nyq}} |e^{-\pi a_{obj}^2 f^2}|^2 DQE(f) df} \quad (2.35)$$

2.4 Lodox Statscan

The Lodox Statscan low dose digital x-ray machine is a linear slot scanning radiography imager capable of taking full body images in 13s. The Statscan utilizes a narrow fan beam that scans the patient by horizontal movement of the C-arm along the Z-direction (Figure 2.5) in synchronism with a narrow slotted image receptor (Gadolinium Oxyphosphate ($Gd_2O_2S:Tb$) scintillator and E2V CCD 256 x 1024 pixels). The slot scanning design of Statscan allows the elimination of the negative effects of scattered radiation without requiring an anti-scatter grid. The result is sharp, high-resolution images with a reduction in radiation dose compared to conventional screen-film and computed radiography systems (Beningfield *et al.*, 2003; deVilliers and deJager, 2003; Irving *et al.*, 2008). These advantages are expected to have potential for use in mammography. Based on the width (256 pixels, 7mm) of the Statscan detector and the definition of slit versus slot scanners given by Lundqvist (2003) and Smith *et al.* (2004), we refer to the Statscan scanner as linear slot scanning radiography (LSSR) throughout this document.



Figure 2.5: The Lodox Statscan machine located in MRC/UCT Medical Imaging Research Unit

Chapter 3

System Requirements for Linear Slot Scanning Mammography

3.1 Introduction

Scanning radiographic geometry provides excellent scatter rejection, high image resolution, and high contrast, with a minimum radiation dose to the patient, by using a narrow fan beam scanned linearly in synchronism with a narrow linear detector (Barnes *et al.*, 1994; Jing *et al.*, 1996; Beningfield *et al.*, 2003; Kroft *et al.*, 2004; Samei *et al.*, 2004).

The system performance of scanned slot mammography depends on the x-ray spectrum and the detector performance. The x-ray spectrum is determined by the x-ray source assembly including the x-ray tube and the filtration. The detector performance depends on the performance of the converting material in the scintillator, and the optical coupling of the generated visible light on the phosphor screen through a fibre optic medium to the CCDs, in which the light photons are converted to an electrical signal. Based on the width (128 pixels, 6mm) of the proposed detector and the definition of slit versus slot scanners given by Lundqvist (2003) and Smith *et al.* (2004), we refer to our scanning system as linear slot scanning mammography (LSSM) throughout the document.

This chapter describes the system design specifications and requirements for a prototype linear slot scanning mammography (LSSM) system and presents the estimated detector

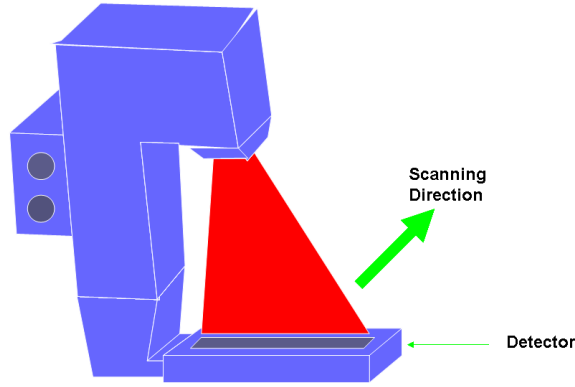
performance. The system was designed to provide a linear scanning geometry with a narrow fan beam to offer good scatter rejection. The detector was selected to provide the high detection efficiency and spatial resolution needed in mammography. The components of the prototype LSSM system were selected based on estimation of the image quality parameters: modulation transfer function (MTF), noise power spectrum (NPS), and detective quantum efficiency (DQE). The image quality parameters were estimated using simulation tools similar to those developed by Siewerdsen *et al.* (2004) but optimized for the LSSM system. The algorithms to estimate the image quality parameters described in Chapter 2 were coded by the author using Matlab (Matlab, The Math-Works, Natick, MA) and were implemented as functions in a graphical user interface (GUI) (see Appendix A for more details).

3.2 System design and requirements

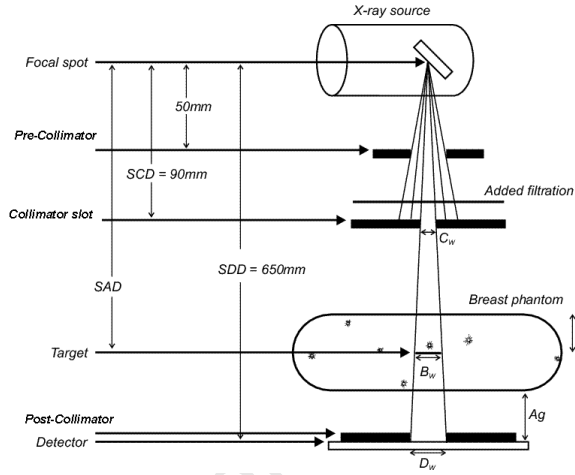
3.2.1 Mechanical design

The LSSM system uses the same linear slot scanning technology as the Lodox Statscan low dose digital x-ray machine (Benningfield *et al.*, 2003) described in Section 2.4. The LSSM system consists of an x-ray tube mounted on a C-arm, utilizing a narrow fan beam that scans the patient by a horizontal movement along the Z-direction in synchronism with a narrow slotted image receptor as shown in Figure 3.1. Two slot-shaped collimators are used. One is mounted on the x-ray tube (collimating slot) to restrict the region of irradiation to a narrow slot, while the second is mounted near the detector to remove scattered radiation. The slot collimator is adjustable to ensure the x-ray beam covers the useful detector area. The required collimation width C_W to produce a beam width of 6mm at the detector surface was calculated to be 0.8mm. The post-collimator has a width of 6mm and depth of 7mm; since the entrance of the detector surface is narrower than the fan beam, and the penumbra is 2mm wider the fan beam, x-ray energy outside the slot is absorbed by the body of the post-collimator. In addition, the post-collimator slot reduces the scattered radiation and the associated noise.

The scanning motion is carried out using a stepped motor with step size of 0.005mm to provide linear continuous motion with a variable scanning speed, depending on the CCD transfer rate (discussed in Section 3.2.4). For minimum geometrical distortion,



(a)



(b)

Figure 3.1: (a) The slot scanning system; (b) geometry of the slot scanning digital mammography system; SDD is the source-to-detector distance, SCD is the source-to-collimator distance, SAD is the source-to-axis distance, C_W is the collimating slot, B_W is the beam width at the patient axis, and Ag is the airgap.

the height of the breast support with respect to the detector (the air gap (Ag)) can be adjusted to maintain the optimum target magnification, namely that which produces the best image resolution.

3.2.2 X-ray tube

In a scanning radiographic system, only a small fraction of the generated x-ray energy is used, which means that the x-ray tube must be operated at the maximum allowable power to achieve the amount of x-ray flux that is needed for breast imaging. The selection of the focal spot size is a tradeoff between the maximum power limit and the spatial resolution. As the power limit increases with increasing focal spot size, the

spatial resolution decreases. Figure 3.2 shows the limiting spatial resolution (5% MTF) calculated for each focal spot size using a simulation model described in Appendix A, and plotted against the focal spot size.

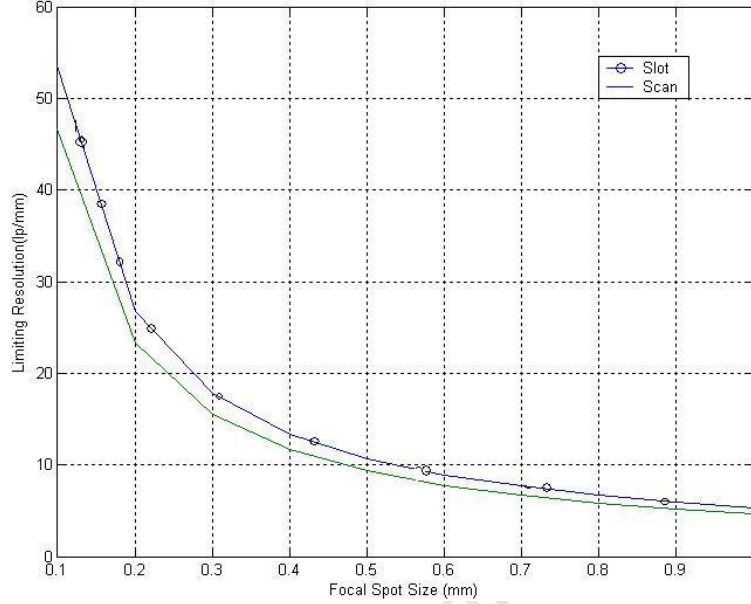


Figure 3.2: The limiting resolution as a function of focal spot size along the length (slot direction) and along the width (scan direction) of the scanning direction.

The optimum focal spot size used in mammography for imaging a low contrast object is in the range between 0.1 and 0.3mm (Bushberg, 2000). Tungsten anode material with a nominal value of 0.3mm has been selected to be used for the prototype system to compromise between the required power loading ($\geq 10\text{kW}$) and a limiting resolution ($> 10\text{lp/mm}$). A Varian x-ray tube RAD70 (0.3mm focal spot) was selected for the prototype mammography system because it has high power rate (15kW) and good resolution ($> 10\text{lp/mm}$). The Varian tube current limits for different kVp values estimated from the RAD70 data sheet (RAD-70, 2004) are illustrated in Figure 3.3.

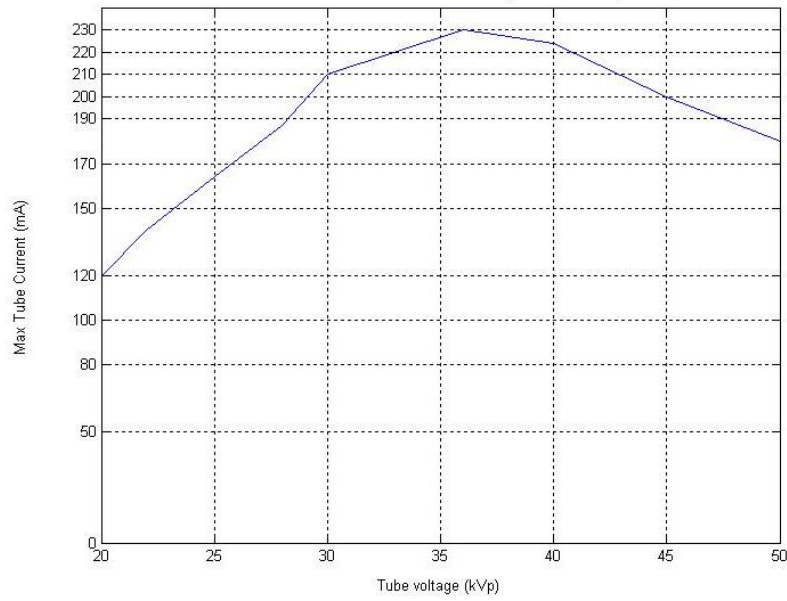


Figure 3.3: Varian tube limits.

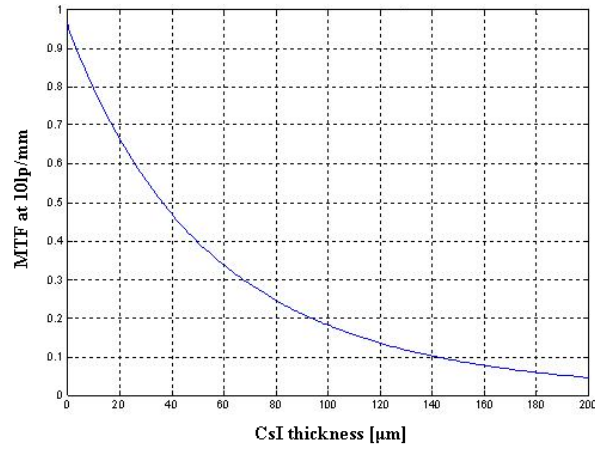
3.2.3 Detector specifications

The detector design and performance depend on the efficiency of the x-ray convertor, which converts the x-ray photons to scintillation light photons, the optical coupling efficiency of the fibre optic plate or taper, and the conversion efficiency of light photons into electrical signal by the CCDs. The basic components of the type of detector proposed for the scanned slot digital mammography system were illustrated in Figure 2.2. A Hamamatsu detector which is described in the following sections has been selected for the prototype system. The detector consists of a high light output Thallium-activated Cesium Iodide (CsI:TI) scintillator that is connected directly through a fibre optic plate to a high resolution and wide dynamic range Hamamatsu CCD operating in TDI mode.

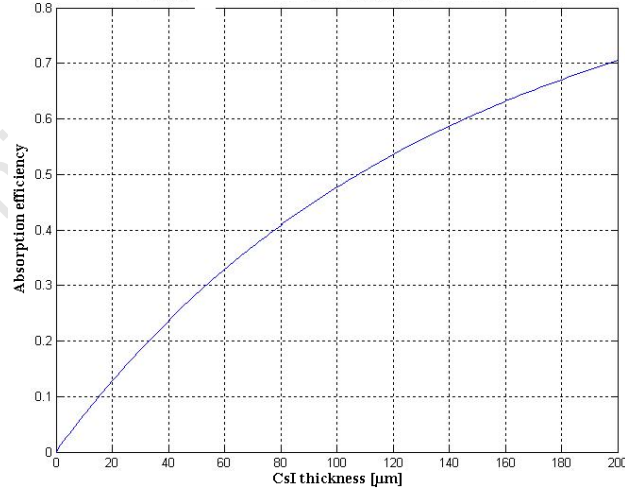
3.2.3.1 Scintillator

The choice of the optimum scintillator depends on the x-ray imaging application, such as mammography or general radiography. In mammography, high absorption efficiency and good resolution are required for visualization of low contrast microcalcification details. The performance of the scintillator depends on the x-ray energy and is determined by the irradiation depth, the amount of light output, and the scintillator material and thickness. The selection of scintillator thickness is a trade-off between the absorption

efficiency and the spatial resolution. Figures 3.4(a) and 3.4(b) show the estimated spatial resolution represented by the MTF and the absorption efficiency of Hamamatsu FOS scintillator (FOS, 1996). The scintillator MTF decreases with increasing thickness, while the absorption efficiency increases with increasing thickness. Using the data provided by Hamamatsu and the simulation tools (described in Appendix A), a $150\mu\text{m}$ thick CsI:TI scintillator with a fibre optic plate (FOS) was found to have a good spatial resolution of more than 10lp/mm and an absorption efficiency of more than 50% at tube voltage of 40kVp.



(a)



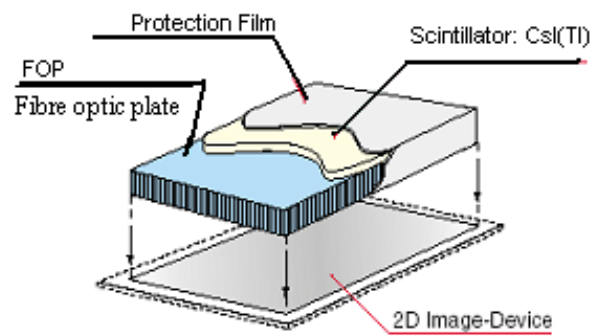
(b)

Figure 3.4: The modulation transfer function (MTF) and absorption efficiency for a high light output (HL) CsI scintillator calculated using *MammoModel* and data provided by Hamamatsu Company (FOS, 1996); (a) the value of MTF at 10lp/mm frequency for each FOS scintillator thickness; (b) absorption efficiency as function of FOS scintillator thickness at 40kVp.

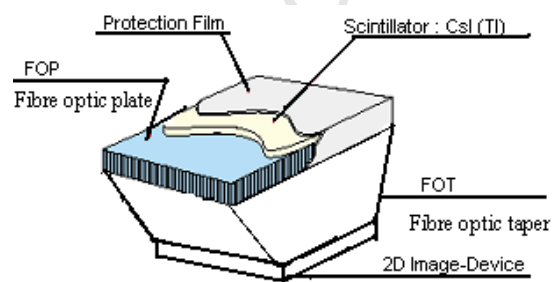
Scintillation light emitted from the scintillator is also a factor affecting the selection of the scintillator thickness. Light output increases as the scintillator thickness increases, so a good balance between light output and the spatial resolution is also required. The generation of light within the scintillator is a random process and depends on the energy of the absorbed photons and the average energy W required to generate a light photon (Zhao *et al.*, 2004; Rowlands and Yorkston, 2000). Hamamatsu has developed a fibre optic plate with a scintillator (FOS, 1996). The estimated value of W for the FOS reported by Zhao *et al.* (2004) is equal to 28 eV.

3.2.3.2 Optical coupling

The collection efficiency of light photons generated in the scintillator is determined by the fraction of light collected onto the charge-coupled device (CCD). The scintillator is either directly coupled to the CCD through a fibre-optic plate as shown in Figure 3.5(a) or optically coupled through a fibre-optic taper in order to reduce the cost of the CCD as shown in Figure 3.5(b). The coupling efficiency of direct coupling is mainly dependent on the transmission efficiency of the fibre-optic plate. For indirect coupling, it is dependent on the transmission efficiency and demagnification of the fibre-optic taper (FOT), in addition to the coupling efficiency of the FOT to the fibre-optic plate (FOP). Direct coupling will offer better detective quantum efficiency (absorption efficiency and resolution) than the indirect coupling and will improve the dynamic range as discussed in the following section.



(a)



(b)

Figure 3.5: Coupling of the scintillator to the CCD (a) direct coupling, and (b) coupling through fibre optic taper (FOT), adapted from (FOS, 1996).

3.2.4 Time delay integration (TDI) mode

The charge-coupled device (CCD) was developed in 1970 (Boyle and Smith, 1970) to replace the vacuum camera tube. It is used for digital radiography because of its high spatial resolution and wide dynamic range. For digital scanning systems, the charge-coupled device is usually operated in time delay integration (TDI) mode, which allows the total charge during the scan to be integrated and shifted down through the column of detector elements to the horizontal readout register as shown in Figure 3.6. Table 3.2.4 shows the technical specifications of the Hamamatsu CCD taken from (CCDS8658, 2004).

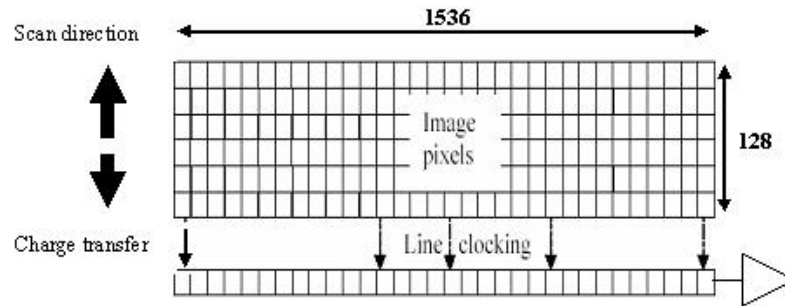


Figure 3.6: Schematic illustration of the TDI mode of operation.

Table 3.1: The technical specifications of the Hamamatsu CCD (CCDS8658, 2004)

	Hamamatsu	Units
Δx	0.048	mm
Δy	0.048	mm
X-ray resolution	6	lp/mm
Quantum efficiency	25%	
Linewidth	1536	pixel
TDI rows	128	pixel
Maximum data transfer rate	1	ms
Dark current	24	$ke^-/\text{pixel/s}$
X-ray sensitive area	221.18×5.94	mm^2
Fill Factor	100%	
Active pixels(3 CCDs)	4608×128	
Full well capacity	1200	ke^-
Readout noise	60	$e^- \text{ rms}$
Dynamic range	20000	

3.2.4.1 Optimum scanning speed

To achieve the maximum system resolution, the mechanical scanning speed must match the charge transfer rate of the CCD. The optimum scanning speed can be estimated from the charge transfer speed. For the Hamamatsu detector, the maximum data transfer rate is 1000 Hz, which is equivalent to 1ms integration time. The calculated charge transfer speed for a $48\mu\text{m}$ pixel was found to be in the range 12 to 48mm/s. This is the optimum scanning speed range.

3.2.4.2 CCD well capacity and dynamic range

The number of electrons generated per pixel, which represent the well capacity and the dynamic range of the detector, were estimated from the technical specifications using *MammoModel* described in Appendix A. The number of electrons generated per pixel depends on the x-ray spectrum, the technique factor (e.g. 40kVp, 200mA, and 0.5mmAl filtration), the quantum efficiency of the scintillator, the scintillator conversion gain, the optical coupling efficiency, and the CCD quantum efficiency. Figure 3.7 shows the estimated number of electrons generated using the Hamamatsu detector for different types of optical coupling, namely direct and indirect coupling. Due to the high coupling efficiency of direct coupling, more electrons are generated with a lower x-ray exposure; this will result in a lower dose to the patient.

Table 3.2: The estimated charge well capacity and the dynamic range for the Hamamatsu detector; technical data are taken from (CCDS8658, 2004).

	Hamamatsu detector
Tube voltage	40kVp
Tube current	200mA
Scanning speed	12 - 48mm/s
Detector exposure	50 - 197 μGy
Total x-ray/pixel	5.3×10^3 - 2.1×10^4
Total coupling efficiency	0.08
Total electron/pixel	1.2×10^6
CCD well capacity required	1.2×10^6
Total readout noise	60
Dynamic range	20000

3.2.5 Summary of system specifications

The system specifications used for building the LSSM system can be summarized as shown in Table 3.3.

Table 3.3: Slot scanning mammography specifications

	Parameter	Hamamatsu
Mechanical design specifications	Type of scan	Linear Slot Scanning Mammography
	Source-to-Detector Distance	650mm
	Source-to-Collimator Distance	90mm
	Source-to-Patient Distance	> 300mm
	Collimating Slot	Adjustable
	Motor Drive	5 μ m step size
	Airgap	Adjustable
	CCD Clocking rate	250-1000Hz
	Scanning speed	12-48mm/s
	X-ray tube	Varian RAD70, Focal spot $\approx$ 0.3mm
	Power rate	15kW
Detector design specifications	Scintillator	CsI:TI-150 μ m
	Optical coupling	Direct coupling
	Number of CCDs	Three (73.728 $\times$ 6.144mm ²)
	Size of detector element	0.048mm
	CCD well capacity required	1.2 $\times$ 10 ⁶
	Dynamic range	20000
	Useful detector size	220 $\times$ 6mm ²
	Dead space	144 μ m
	Image matrix size	5000 $\times$ 4608
	Gray levels	4096 (16bits)
	Image size	$\approx$ 46MByte
	USB data transfer rate	> 300Mbit/s

3.3 Modelling the performance of LSSM

3.3.1 Energy spectrum simulation

The energy spectrum of the slot scanning beam is characterized by the tube settings (tube voltage kVp and tube current mA), the effective beam width (B_W), and the

scanning speed (S_{Speed}). The photon fluence in $photons/mm^2$, modified for the slot scanning beam for a given tube current and effective scanning time (mAs), can be written as:

$$q_{Slot}(E) = q_0(E)mA\frac{B_W}{S_{Speed}} \quad (3.1)$$

$q_0(E)$ is photon fluence, which has units of $photons/mm^2$ per mAs , taken from published data (Boone, 1997) and calibrated for the slot scanning beam using a measured exposure at known tube voltage, tube current, beam width and scanning speed. Figure 3.8 shows the calculated spectrum for the given technique factors.

The exposure in (μGy) at the surface of the phantom can be estimated from the calculated photon fluence in $photons/mm^2$ modified for the slot scanning beam and the x-ray fluence per unit exposure $\xi(E)$:

$$E_{Expo} = \int_{E_{min}}^{E_{max}} q_{Slot}(E) \times \xi(E)^{-1} dE \times 8.76 \quad (3.2)$$

The factor of 8.76 converts from mR to air kerma in μGy .

The x-ray fluence per unit exposure ($photon/mm^2/milliRontgen$), $\xi(E)$ can be estimated using the empirical equation (Boone, 2000):

$$\xi(E) = [x + y\sqrt{E}\ln(E) + \frac{z}{E^2}]^{-1} \quad (3.3)$$

where x , y , and z are constants equal to $-5.023290717769674e-06$, $1.810595449064631e-07$, and 0.008838658459816926 respectively. The beam width in the scan direction increases linearly with the distance from the source. As the slot moves with constant speed, the time taken for the slot to cross any position in the beam increases linearly with distance from the source, and is given by the relation:

$$t = \frac{r}{v_{Speed}} \quad (3.4)$$

Where r is distance from the source; v_{Speed} is the constant scanning speed. As the exposure is proportional to the product of the scanning time and photon flux, the exposure in the scanning direction is proportional to :

$$t = \frac{r}{r^2} = \frac{1}{r} \quad (3.5)$$

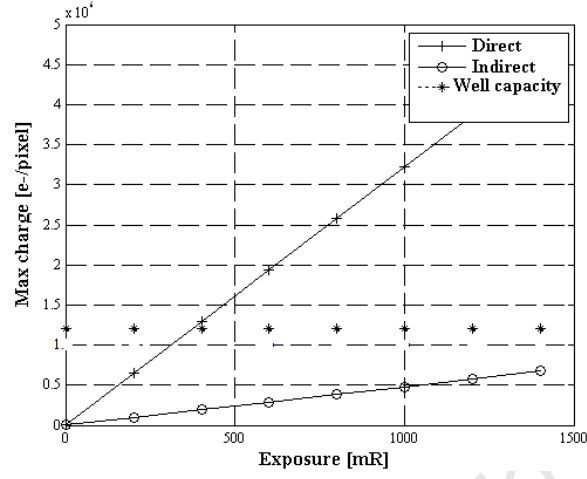


Figure 3.7: The estimated maximum charge capacity for a direct and indirect coupling detectors using the Hamamatsu technical specifications and *MammoModel* described in Appendix A.

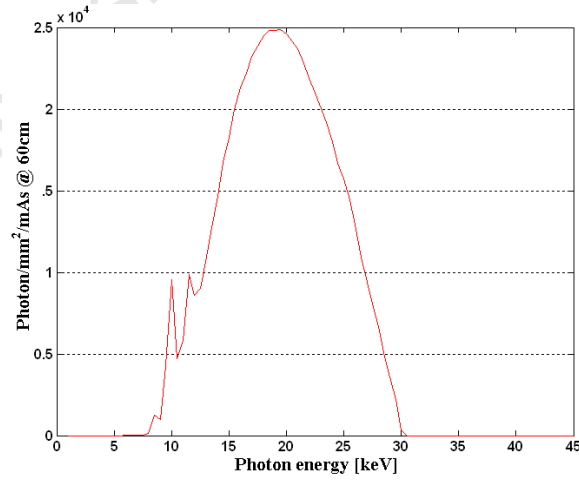


Figure 3.8: The calculated tungsten energy spectrum at a distance of 60cm, tube voltage of 30kVp, tube current of 100mA, and external filtration of 0.5mm Al.

Hence the doses were corrected for source to detector distance for linear slot scanning, following a $1/r$ rather than a $1/r^2$ relation as explained also in detail in (deVilliers and deJager, 2003; Irving *et al.*, 2008).

This relation is confirmed experimentally in Chapter 4. The detector Air Kerma D_{Expo} was then estimated using the following equation:

$$D_{Expo} = \frac{SPD}{SDD} \int_{E_{min}}^{E_{max}} q_{Slot}(E) e^{-\mu_B x} \times \xi(E)^{-1} dE \times 8.76 \quad (3.6)$$

where SDD is source to detector distance; SPD is source to phantom distance; μ_B is the energy dependent linear attenuation coefficient of the breast tissue; and x is the thickness of the breast.

3.3.2 Imaging task

The imaging task is applied to an object consisting of a microcalcification (calcium hydroxyapatite) of density 3.3g/cm^3 embedded in a semicircular cross-section of breast tissue (50% glandular and 50% adipose) of radius 8.5cm and thickness of 4cm. The attenuation coefficients of breast tissue and microcalcifications were computed using the program Xcom (Berger *et al.*, 2007). Figure 3.9 shows the calculated filtered spectrum and the attenuated spectrum for different imaging tasks. The attenuated energy spectrum through the breast and the microcalcifications were calculated using the Lambert-Beers law (Boone, 2000):

$$q_{Breast}(E) = q_{Slot} e^{-\mu_B X_B} \quad (3.7)$$

$$q_{MicroCal}(E) = q_{Slot} e^{(-\mu_B(X_B - x_C) - \mu_C x_C)} \quad (3.8)$$

3.3.3 Scatter

Detected scatter photons degrade image quality by reducing image contrast and increasing the image noise. The reduction in image contrast is characterized by the scatter fraction, which is equal to the ratio of the detected scatter photons to the total detected signal (primary and scattered photons).

Several techniques have been developed to reduce the contribution of scattered photons to the image, for example the antiscatter grid and air gap techniques (Barnes *et al.*, 1980; Sorenson *et al.*, 1980; Sorenson and Floch, 1985; Junck *et al.*, 1996). The antiscatter grid result in a 60% reduction in the scatter-to-primary ratio (SPR) (Junck *et al.*, 1996), while the air gap of 3cm can reduce the SPR by a factor of 3 (Jing *et al.*, 1998).

The most effective technique used for reducing the contribution of scatter photon to the image is the slot-scanning technique (Barnes *et al.*, 1977; Jing *et al.*, 1998; Samei *et al.*, 2004, 2005). In full-field digital radiography systems the use of slot-scanning reduces the scatter fraction by a factor of two to three compared to other techniques (Samei *et al.*, 2005).

Several techniques have been developed for estimation of SPR in digital mammography using Monte Carlo simulations (Jing *et al.*, 1998; Boone *et al.*, 2000). Jing *et al.* (1998) quantified the amount of scatter in slot scanning for digital mammography; the scatter-to-primary ratio for a breast phantom of thickness 2-6cm ranged from 0.1 to 0.17 for slot width 4mm. Boone *et al.* (2000) investigated SPR for full field and slot scanning digital mammography. They developed an empirical equation for estimation of SPR for full field digital mammography:

$$SPR = [a + b \times D^{-1.5} + c \times thickness^{-0.5}]^{-1} \quad (3.9)$$

Where D is breast diameter in cm; $a = 2.35452817439093$, $b = 22.3960980055927$, $c = 8.85064260299289$

Using Monte Carlo simulation, the SPR was calculated as a function of slot width (Boone *et al.*, 2000). The estimated scatter reduction factors (ratio of SPR of full field to slot scanning) for an average breast thickness of 4cm, and slot width of 2mm was estimated to be 11.7; this factor drops to 3.7 for a 10 mm slot width. These results were consistent with the simulation results of Jing *et al.* (1998).

Another reason for the low SPR of the slot scanning technique, in addition to the slot width, which allows the use of a smaller detector area than in full field imaging, is the low angular acceptance of the post-collimator as shown in Figure 3.10.

The SPR is directly proportional to the slot width (w) and inversely proportional to the depth of the post-collimator slot.

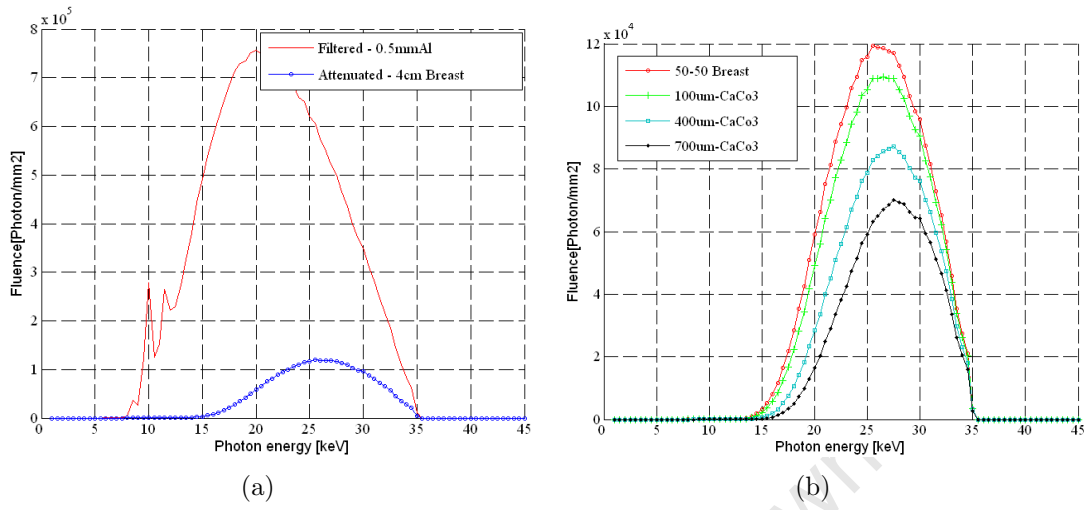


Figure 3.9: The calculated energy spectrum modified for slot scanning; (a) the filtered spectrum (0.5mmAl, 6mm beam width, 35kVp and 200mA at 60cm from the source) and attenuated spectrum through 4cm breast (50:50 glandular/adipose tissue); (b) calculated spectrum for different imaging tasks, 50:50 breast and different microcalcifications (100, 400, and 700 μ m).

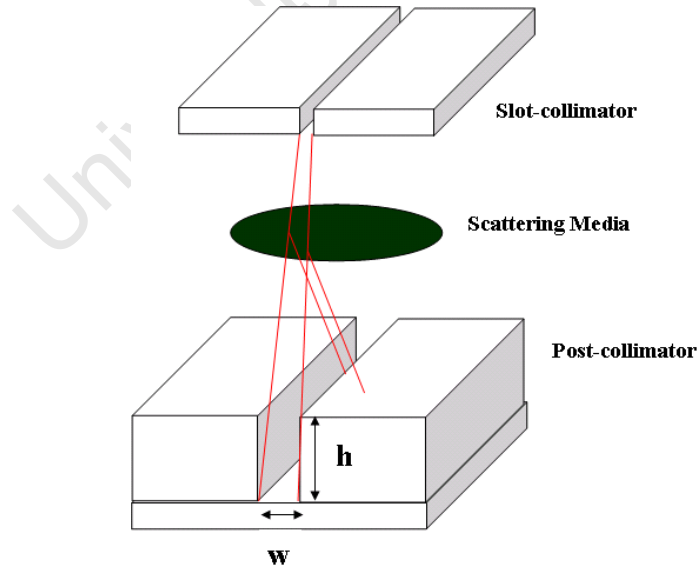


Figure 3.10: The scatter effect in slot scanning.

$$SPR \propto \frac{w}{h} \quad (3.10)$$

Using the data provided by Jing *et al.* (1998), and Boone *et al.* (2000), the SPR for the proposed slot scanning mammography system was estimated using a fitted equation for different breast thicknesses and kVp values (Figures 3.11 and 3.12). The equations were coded by the author using Matlab (Matlab, The MathWorks, Natick, MA) and were implemented as functions in a graphical user interface (GUI) (see Appendix A for more details).

$$SPR_{slot} = ax^3 - bx^2 + cx + d \quad (3.11)$$

Where x is breast thickness. The value of the parameters a, b, c, and d for each breast thickness were given in Appendix A.

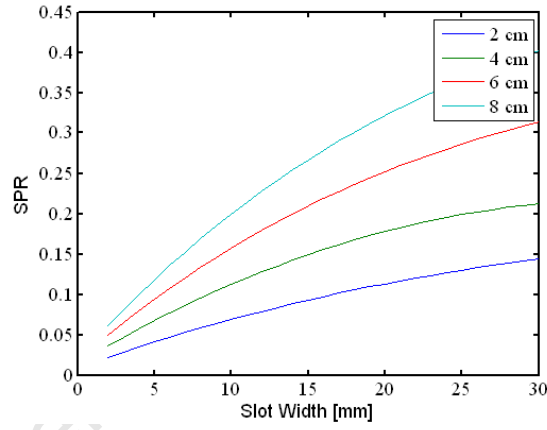


Figure 3.11: The effect of slot width in scatter rejection for different breast thickness.

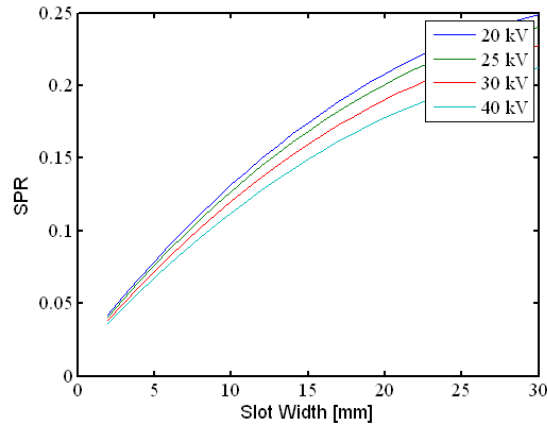


Figure 3.12: The effect of slot width in scatter rejection for different kVp.

3.3.4 System resolution MTF

The presampling MTF of the proposed slot scanning mammography system was estimated from the scintillator MTF, the focal spot MTF, and the blurring due to the integration of quanta in the CCD pixel. The focal spot MTF was determined using the estimated intensity distribution of the focal spot along the focal length (slot direction), and along the focal width (scan direction) as shown in Figure 3.13. The estimated intensity distribution was simulated using the measured focal spot intensity distribution data of the Varian x-ray tube (RAD 70) provided by the company. The focal spot MTF was calculated from the line spread function, which represents the intensity distribution of the focal spot. To account for various combinations of magnification and focal spot size, the MTF was scaled with $(m-1)d$, where m is the target magnification, and d is the focal spot size. Figure 3.14 shows the effect of using a Thallium-activated Cesium Iodide (CsI:TI) scintillator rather than Gadolinium Oxyphosphate (GdOS) on the detector resolution (MTF). Figure 3.14 shows the effect of different MTF components contributing to the final system MTF. The MTF spatial frequencies above the Nyquist frequency might be aliased and degrade the imaging performance of the LSSM system. However, the rapid drop of the pre-sampling MTF above the Nyquist frequency acts as an anti-aliasing filter.

3.3.5 Noise power Spectrum NPS

Using Equation 2.16, the noise power spectrum was estimated for the LSSM detector as a function of detector exposure and electronic noise. Then the NPS was normalised by the overall mean signal of the detector to give the normalised noise power spectrum (NNPS), as shown in Figure 3.16 and Figure 3.17.

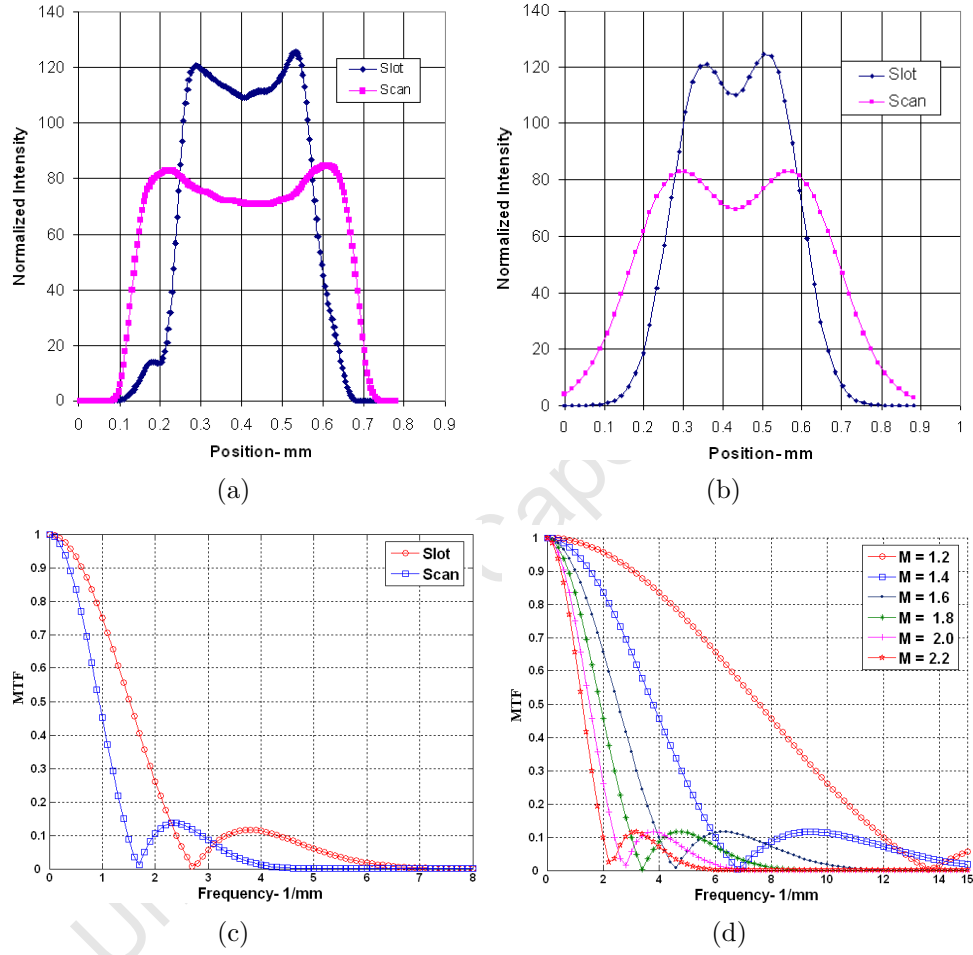


Figure 3.13: The estimated focal spot MTF using the measured intensity distribution of the Varian x-ray tube (RAD 70, 0.3mm focal spot)(RAD-70, 2004): (a) measured intensity distribution in direction along focal length (slot direction) and along width (scan direction); (b) simulated intensity distribution; (c) calculated MTF in slot and scan direction; (d) calculated focal spot MTF for different object magnifications (m), magnification $M=2$ means object midway between detector and source.

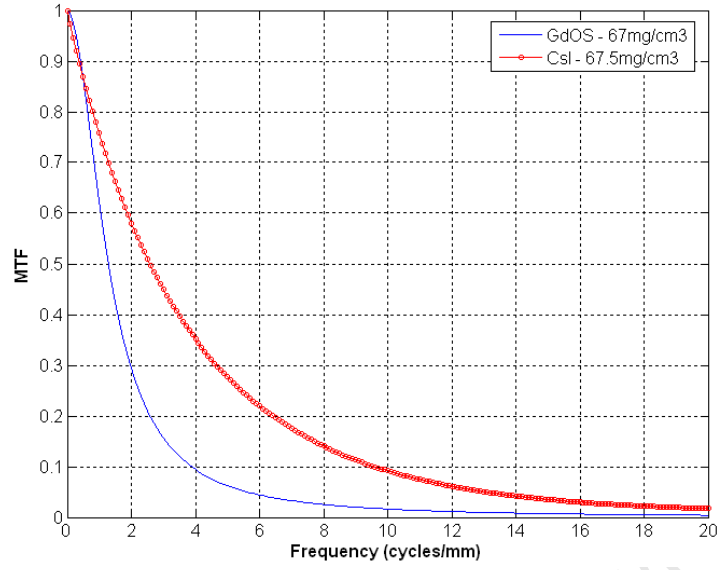


Figure 3.14: The scintillator MTF.

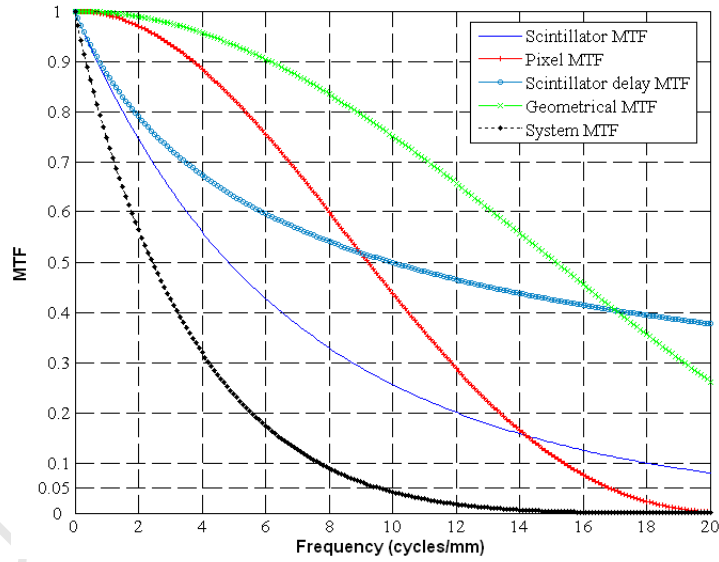


Figure 3.15: The estimated system MTF for LSSM system using a direct coupling detector consisting of the Hamamatsu FOS and CCDs.

3.3.6 Detective quantum efficiency DQE

The detective quantum efficiency describes the efficiency of an imaging system, combining the effect of detector x-ray absorption efficiency, the spatial resolution, and the detector noise in one parameter. The absorption efficiency and the detector noise are characterized by the detective quantum efficiency at zero spatial frequency ($DQE(0)$), while the limiting spatial resolution is described as the frequency at which MTF falls to 5% of its value. Using the estimated MTF, NNPS, and Equation 2.7, the detec-

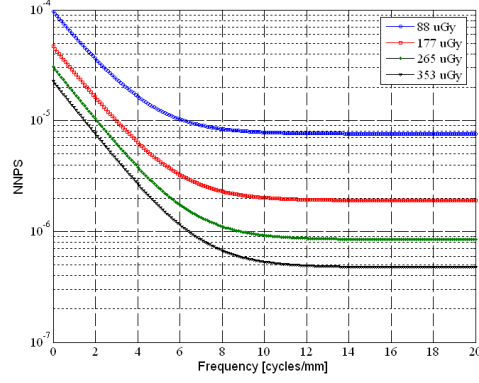


Figure 3.16: The normalized noise power spectrum (NNPS) for the LSSM detector; data are shown for three detector exposures.

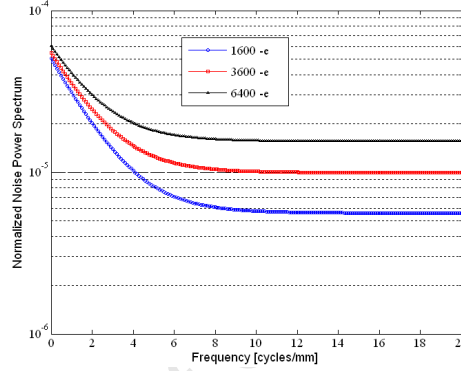


Figure 3.17: The normalized noise power spectrum (NNPS) for the LSSM detector; data are shown for different levels of electronic noise.

tive quantum efficiency was estimated for LSSM system as shown in Figure 3.18 and 3.19 as a function of scintillator thickness and detector coupling efficiency, respectively. As seen in Figure 3.18, the $150\mu\text{m}$ CsI scintillator shows a good balance between the frequency-dependant detective quantum efficiency ($\text{DQE}(f)$) and the spatial resolution. The $150\mu\text{m}$ scintillator shows a detective quantum efficiency at zero spatial frequency ($\text{DQE}(0)$) of more than 0.5 (50%) combined with a good resolution of nearly 10lp/mm before the DQE falls to zero. As expected, the DQE is lower for lower detector coupling efficiency due to the secondary quantum sink that results from the poor coupling efficiency.

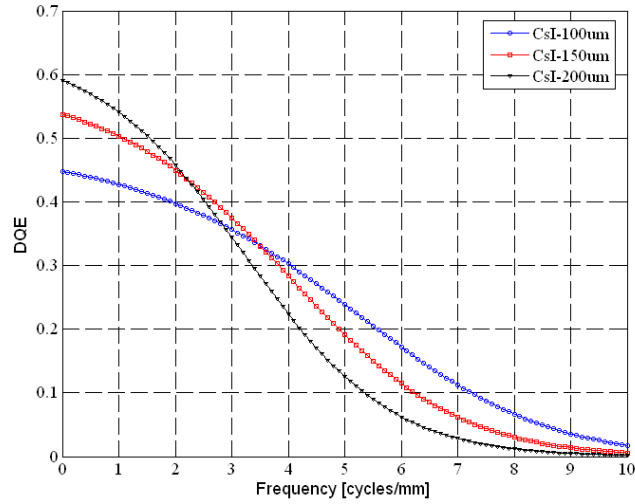


Figure 3.18: The detective quantum efficiency (DQE) as a function of scintillator thickness.

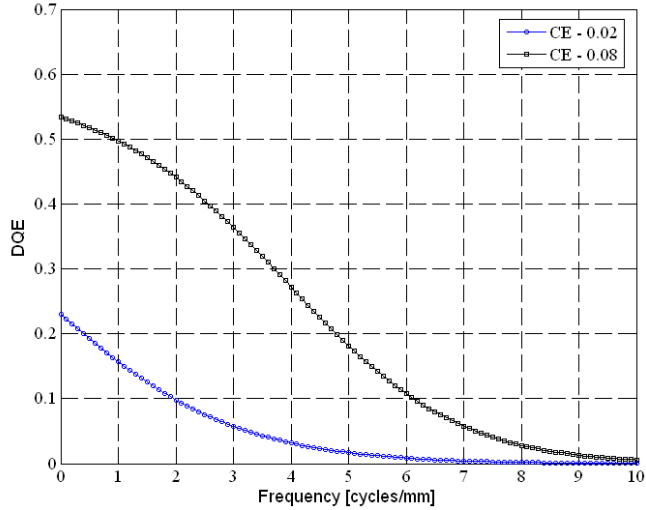


Figure 3.19: The detective quantum efficiency (DQE) as a function of coupling efficiency.

3.4 Summary and discussion

The Lodox Statscan linear slot scanning configuration was used to develop a low dose high resolution digital system for mammography applications. The system requirements and design specifications of linear slot scanning mammography (LSSM) were derived using a computational tool, titled *MammoModel* developed by the author and described in Appendix A. The selection of the x-ray tube and detector components was based on the mammography system requirements. The Varian x-ray tube with 0.3mm focal spot at 15kW provided the required power and resolution for mammography applications. A high performance detector consisting of a FOS scintillator (e.g CsI:TI - 150 μ m) which is

directly coupled to a CCD (e.g Pixel size - $50\mu\text{m}$) operating in TDI mode will provide a good resolution of nearly 10lp/mm and detection efficiency of more than 50% which is enough to detect microcalcifications of $100\mu\text{m}$. The estimated DQE which was obtained from the presampling MTF, the detector gain G , the average number of input X-ray quanta Q and the digital noise power spectrum NPS of the detector using *MammoModel*, showed that the LSSM has good performance (DQE of more than 50% at a spatial frequency of 0lp/mm) as shown in Figure 3.18. In comparison, the DQE of screen film detectors is typically about 0.2 to 0.3 at zero spatial frequency (Dobbins *et al.*, 1995). As shown in Figure 3.18, the $150\mu\text{m}$ FOS scintillator thickness showed a good balance between the detection efficiency and the spatial resolution. All the data presented in this chapter were generated using *MammoModel*. The following chapter presents the LSSM system which was assembled from the selected components and the system specifications derived in this chapter.

3.5 Author's contribution

The author developed a computational tool *MammoModel* (described in Appendix A) for simulating the performance of linear slot scanning digital mammography based on his analysis of the relevant theory. This computational tool was used to derive the system requirements and design specifications of a prototype linear slot scanning mammography (LSSM) system.

Chapter 4

Linear Slot Scanning Mammography System Evaluation

This chapter presents a brief description and an evaluation of the prototype linear slot scanning mammography system developed at the MRC/UCT Medical Imaging Research Unit, University of Cape Town. The selection of components for the system was based on simulation of the image quality parameters and requirements of low dose, high resolution, and better contrast digital mammography as described in Chapter 3. This system was built by members of the MRC/UCT Medical Imaging Research Unit.

A tissue-equivalent phantom designed for mammography (Model 011A) manufactured by CIRS (Computerized Imaging Reference Systems, Norfolk Virginia, USA) was used for system evaluation. The performance of LSSM was also investigated in terms of efficiency, resolution, and detector noise using image quality parameters such as modulation transfer function (MTF), noise power spectrum (NPS), detective quantum efficiency (DQE), and contrast-to-noise ratio (CNR).

4.1 System description and specifications

4.1.1 System description

The linear slot scanning mammography x-ray prototype system (LSSM) consists of an x-ray source which is mounted on a C-arm with a Hamamatsu detector; the C-arm can be moved relative to a table by means of a stepper motor and linear drive. The different subsystems all interface through a USB and an RS232 adapter to a personal computer which serves as the controller and allows the image data to be saved. Figure 4.1 shows the system configuration of the LSSM system.

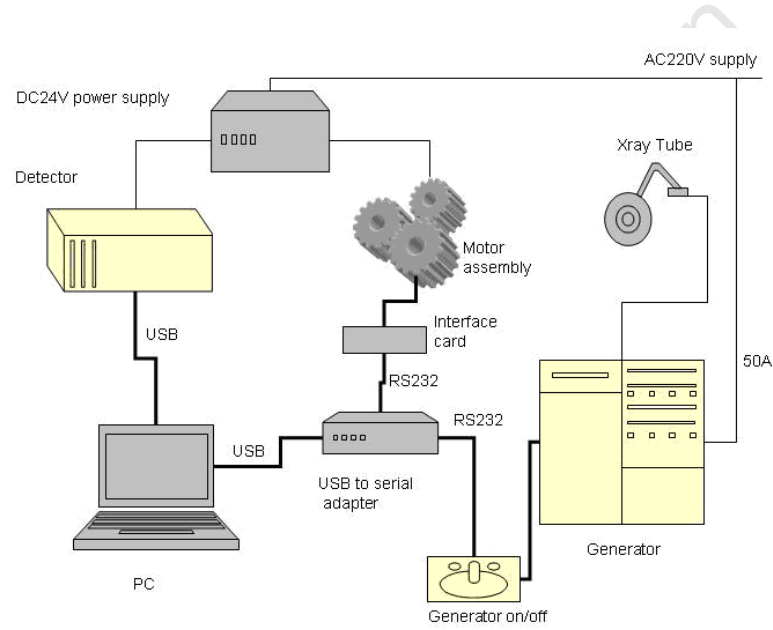


Figure 4.1: The LSSM system configuration.

4.1.2 System specifications

Figure 4.2 and Table 4.1 show the LSSM system and the specifications of the main components of the system.

Table 4.1: Summary of LSSM specifications; components are shown in Figure 4.2.

Component	Manufacturer	Specifications
1- X-ray source	Varian	Tube Voltage: 20 - 49 kVp Tube current: 5 - 200mA Focal spot size: 0.3mm
2- X-ray collimation		Beam shaping collimator: WxL 1x10mm ² Slot collimator: WxL (0.2 - 0.8)x40mm ² Post collimator: WxLxH 6x200x0.8mm ³
3- X-ray detector	X-ray technology(X-tek)	Hamamatsu CCD and FOS
4- Breast Phantom		

4.1.3 X-ray source and collimation

A Varian x-ray tube RAD70 (0.3mm focal spot) was used to provide the required power and resolution for linear slot scanning mammography as described in Chapter 3. The x-ray beam was collimated using three stages of collimation (see Appendix B) ; an initial collimator was used for shaping the scanning beam, a slot collimator was used to produce a fan beam, and a post collimator situated on the top of the detector was used for scatter rejection.

4.1.4 LSSM detector

The LSSM detector, consisting of a Hamamatsu FOS (FOS, 1996) and a Hamamatsu CCD (CCDS8658, 2004), was incorporated in an enclosure with control electronics manufactured by X-ray technologies (X-Tek, 2006) for industrial applications. The area image sensor consists of three sensor modules of 1536 horizontal pixels each, giving a total of 4608 horizontal pixels. The image sensor has 128 vertical pixels operating in a time delay integration (TDI) mode. This means that the sensor should be scanned in the scan direction at a speed equal to clocking of the image data from one vertical line to the next. The two-phase nature of the CCD detector limits the direction of scanning for TDI purposes to one direction only, which implies that the detector will need to return to the start position before a scan is repeated. The software allows for the motor speed, acceleration and distance moved, to be set in conjunction with the clocking speed on the CCD detector and the duration of x-ray illumination from the source, in order to optimize these parameters during experiments.

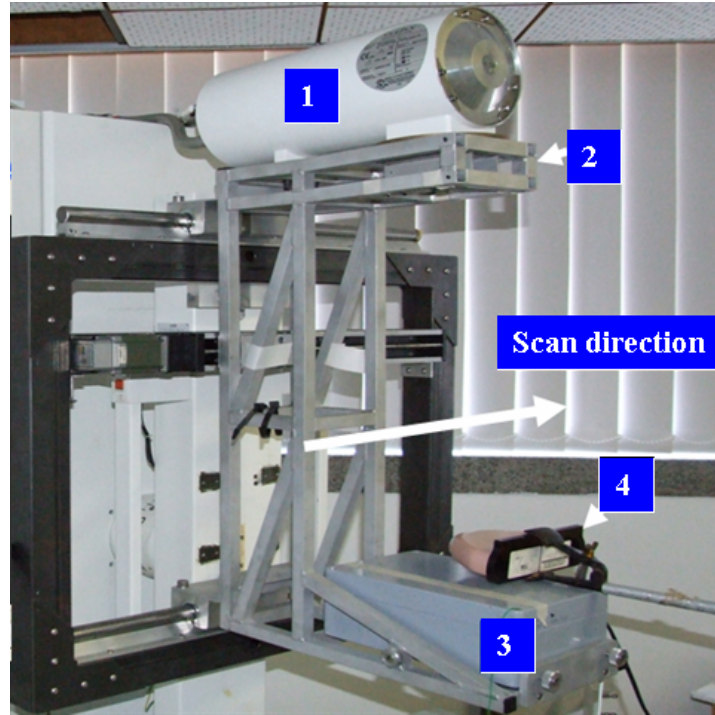


Figure 4.2: Slot scanning mammography system (LSSM); numbered components are described in Table 4.1.

4.1.5 Operator user interface and image acquisition

The operator interface allows the user to set configuration parameters for the system, perform scanning procedures, and save image data. The image acquired from the detector consists of a stream of image data of 4608 horizontal pixels and selected vertical pixels in the scanning direction. The acquired image has 16 bits (65536 levels) of grayscale discrimination per pixel and a file size of approximately 46MBytes (4608 x 5000 pixels). The image data are sent to the computer through a USB cable.

4.2 System evaluation

Several tests and investigations were carried out to evaluate the system performance of the prototype LSMM system. Beam and image quality parameters such as half value layer (HVL), modulation transfer function (MTF), noise power spectrum (NPS) and detective quantum efficiency (DQE) were evaluated and dose measurements were made. A further evaluation was carried out using a tissue-equivalent phantom designed for

mammography, for comparison of the performance of the LSSM system with current mammography systems.

4.2.1 Spectrum calibration and dose measurements

The x-ray energy spectrum of the LSSM system x-ray tube was estimated as described in Chapter 3, and then calibrated using the entrance exposures at a distance of 60cm from the source measured using a relatively large (60cm^3) flat ionisation chamber (Radcal 10X5-60E with Radcal 9010 dosimeter). The ionisation chamber was placed about 60 mm from the detector to minimize backscatter effects. The flat chamber was scanned linearly with optimum scanning speed matching the charge transfer rate of the CCD. Figure 4.3 shows an example of the calculated energy spectrum used for the estimation of the image quality parameters (modulation transfer function (MTF), noise power spectrum (NPS), and detective quantum efficiency (DQE)) for LSSM. The mean glandular dose for different glandular tissue levels and different breast thicknesses were estimated using the calculated energy spectrum and the fit-equation developed by Boone (2002). Figure 4.4 shows the estimated mean glandular dose as a function of the measured entrance exposure at a distance of 60cm from the source using a tube voltage of 30kVp and different tube currents. Figure 4.5 shows the estimated exposure for different tube volt-

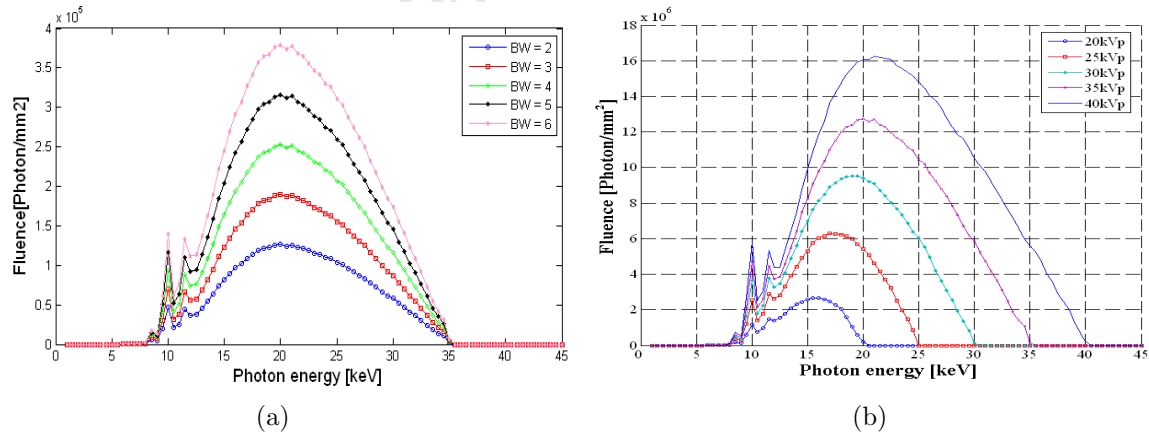


Figure 4.3: The calculated energy spectrum modified for slot scanning (a) different beam widths (BW), 35kVp and 200mA (b) different tube voltages, beam width 6mm and 200mA.

ages using Equation 3.2 compared to the values measured using the ionisation chamber. The ionisation chamber was placed at a distance of 60cm from the source. Dose was then corrected for source to detector distance for linear slot scanning, following a $1/r$

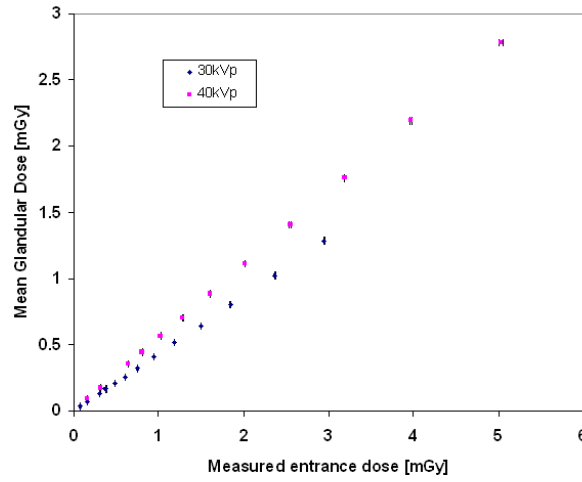


Figure 4.4: The mean glandular dose MGD as function of measured entrance dose at the surface of the phantom.

rather than a $1/r^2$ law (Section 3.3.1; (deVilliers and deJager, 2003; Irving *et al.*, 2008)). This relation is confirmed experimentally for the LSSM system as shown in Figure 4.6. The calculated exposure using the inverse distance law shows a good agreement (with $< 5\%$ error) for the distance range between 50 and 65cm from the source where the breast is normally placed in mammography. The estimated and the measured exposure show a good correlation of 0.9999, and slope of nearly 1 for tube voltage 35kVp as indicated in Figure 4.5.

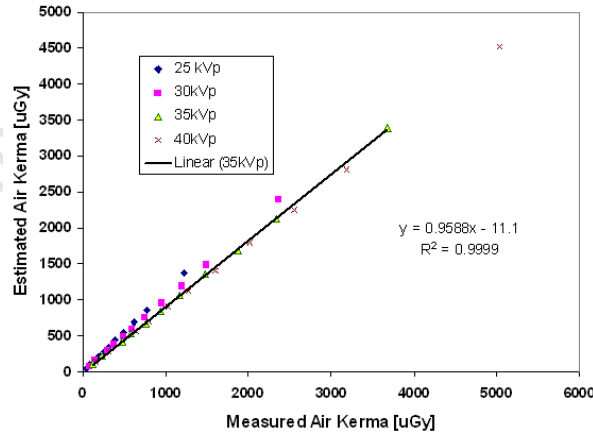


Figure 4.5: The estimated and measured entrance exposure at a distance of 60cm from the source at different tube voltages, with a correlation of 0.9999, and slope of 0.96 for 35kVp.

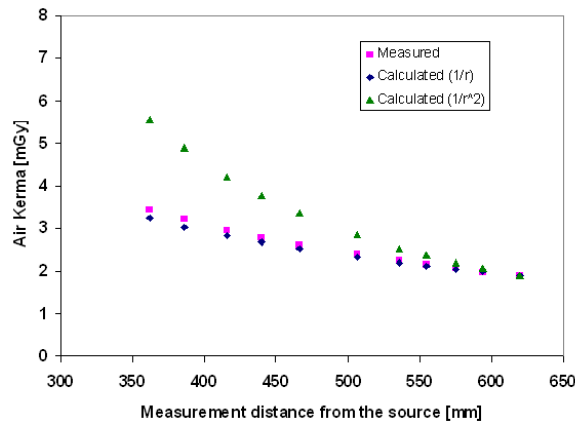


Figure 4.6: The measured entrance exposure as a function of distance from the x-ray source and the estimated dose using the inverse distance ($1/r$) and the inverse square distance ($1/r^2$) law.

4.2.2 Beam quality measurement

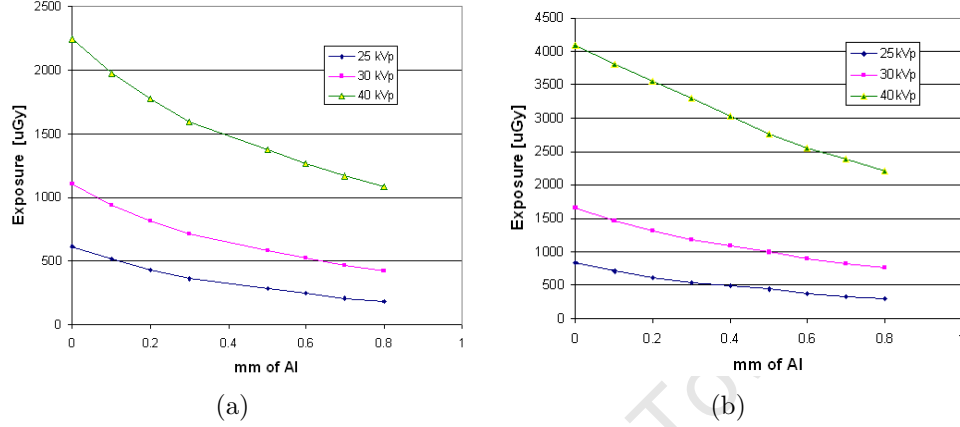
The half value layer (HVL) is the thickness of material (e.g aluminium (Al)) needed to reduce the x-ray exposure to half of its unfiltered value. The HVL measures the quality of the x-ray beam in terms of its penetrability, which is greater at higher energies. The penetrability of an x-ray beam also varies for different systems because it is dependent upon the components that generate the x-ray spectrum.

The HVL of the LSSM x-ray beam was measured using a relatively large (60cm^3) flat ionisation chamber (Radcal 10X5-60E with Radcal 9010 dosimeter) and the Lodox half value layer filter set. The ionisation chamber was placed about 60 mm from the detector to minimize backscatter effects. The filters were positioned half-way above the detector to prevent scattered radiation from reaching the detector. This was also far enough from the source to ensure adequate coverage with such a narrow beam width. Measurements were taken during linear scanning of the flat chamber with optimum scanning speed, matching the charge transfer rate of the CCD.

Figure 4.7 shows that the half value layer increases with increasing x-ray energy due to the high penetrability at higher energies. The measured HVL for a standard mammography procedure was found in the range between 0.4 to 0.9mm Al for tube voltage 25 to 40kVp and tube current 100mA; the typical mammography tube currents range from 80-100mA (Huda and Slone, 1995). These values were found to be slightly higher than reported elsewhere for conventional mammography (Boone, 2002).

Table 4.2: The half value layer for different tube voltages.

Tube voltage	0.5mm Al	0.9mm Al
25	0.47	0.54
30	0.55	0.7
40	0.72	0.91

**Figure 4.7:** The Half Value Layer of the LSSM x-ray beam using external filtration of (a) 0.5mm Al; (b) 0.9mm Al

4.2.3 Detector response

Figure 4.8 shows the detector response with exposure of the phantom over the range 0.3 to 3.0mGy. As expected for unprocessed images, a linear relationship between air kerma and detector exposure was observed ($r = 0.9994$). The dark signal associated with dark current in the CCD, which depends on the scanning speed and the total integration time of the CCD, is added to the detector signal. The additive noise (dark noise, readout noise, digitization noise) can be estimated for the fitted line (digital units (DU) value at 0 signal level) and was found to be 10374 DU. The measured dark signal (mean pixel value of an image taken without x-ray exposure) was found to be 10335 DU. This value agreed (within 99%) with the estimated value from the fitted line. The dark noise in the CCD σ^2 (electrons), associated with dark current, can be estimated from the CCD dark current, the pixel sensitive size, and the total integration time of the time delay integration (TDI).

$$\sigma_D^2 = Ds \times Pixz \times T_{TDI} \quad (4.1)$$

Ds is the dark current in $e^-/pixel/s$; $Pixz$ is the CCD sensitive pixel size; T_{TDI} is total integration time, which can be given by the number of TDI raw divided by the scan speed. Equation 4.2 shows how dark noise depends on the scanning time and the TDI

process.

$$\sigma_D^2 = \frac{Ds \times TDI_{raw} \times Pixz}{S_{Speed}} \quad (4.2)$$

S_{Speed} is scan speed in mm/s.

The other additive noise such like electronic readout noise and digitization noise from the analog-to-digital converters (ADCs) can be estimated from Hammamatsu detector data sheet.

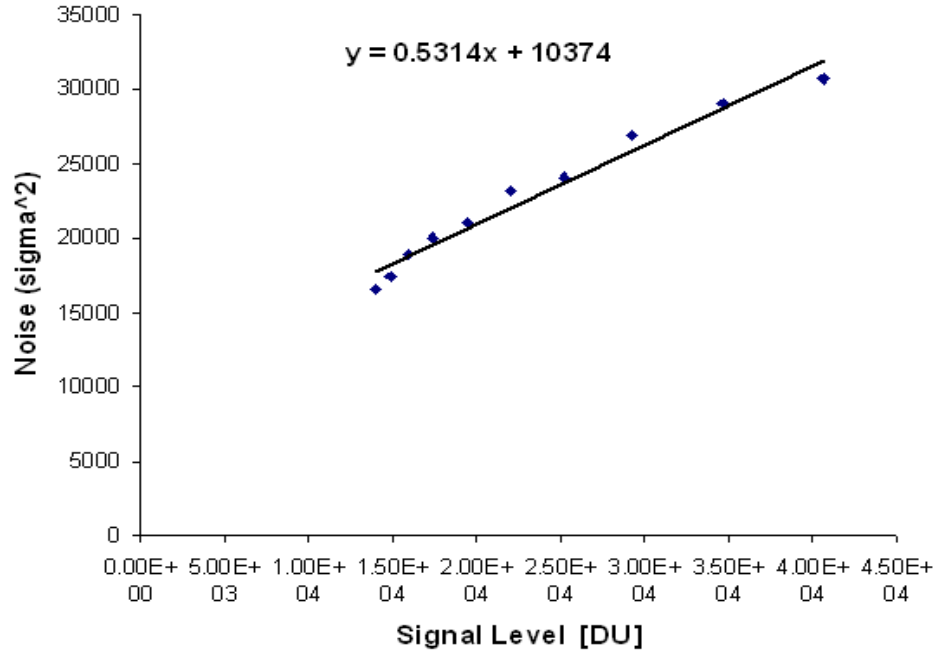


Figure 4.8: Signal level in DU (digital units) as a function of square of standard deviation(noise level).

4.2.4 Image quality measurements

The image quality parameters MTF, NPS, and DQE of the LSSM detector were measured at different exposure levels. The algorithms to calculate MTF, NPS, and DQE were coded using Matlab (Matlab, The MathWorks, Natick, MA).

4.2.4.1 Modulation transfer function

The MTF was measured using an edge detection method described by deVilliers and deJager (2003) and Samei *et al.* (2004). The edge test device consists of a tungsten

plate ($3 \times 3 \times 0.3 \text{ mm}^3$) placed on top of a 4cm thick tissue-equivalent material. The tissue-equivalent material was placed 15mm from the detector to account for the degradation due to the finite focal spot size. Ten images were acquired using LSSM at technique factors of 30kVp and 125mA at a distance close to the detector face. Using the images of the edge device, the MTF in the slot and scan directions was calculated using the following method as described by deVilliers and deJager (2003):

- The edge spread function (ESF) was obtained from 500 projected horizontal lines parallel to the edge.
- The projected lines were corrected for nonuniformity and noise by averaging mean pixel value of the projected lines.
- The line spread function was obtained from the derivative of the smoothed edge spread function (ESF).
- The MTF was computed from the absolute magnitude of the Fast Fourier Transform of the line spread function (LSF).
- The MTF values computed for each acquired image were then averaged to reduce the effect of noise.

This procedure is captured in the Matlab file *Edge-MTF-Calculator.m* on the accompanying CD.

Figures 4.9 and 4.10 show the measured edge spread function and the line spread function in the slot direction using the edge detection method. Figure 4.11 shows the measured MTF in the slot and scan directions. The MTF in the scan direction is slightly less than that in the slot direction; this agrees with results found using the line pair tools to be discussed later in this chapter.

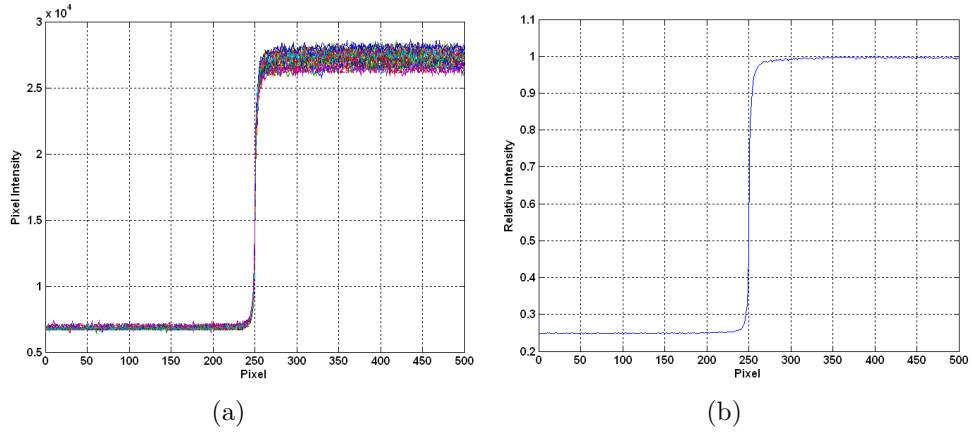


Figure 4.9: The measured edge spread function(ESF); (a) the noisy ESF (b) the average ESF.

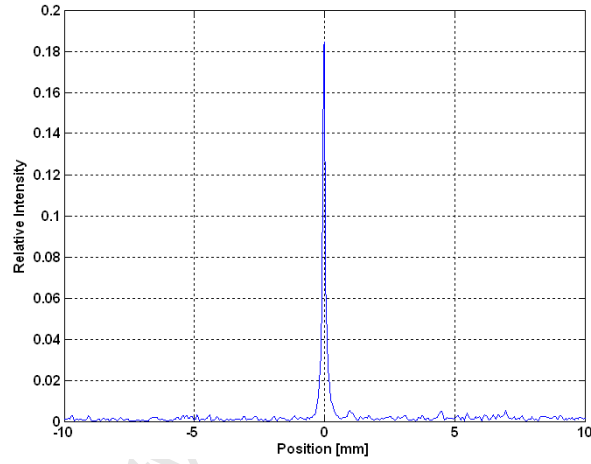


Figure 4.10: The measured line spread function (LSF).

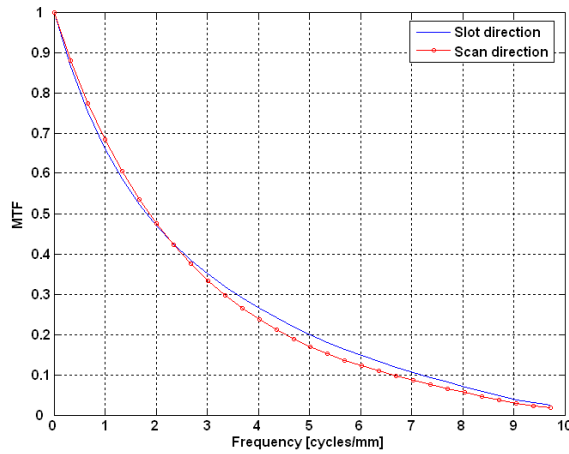


Figure 4.11: The measured modulation transfer function (MTF) in the slot and scan directions.

4.2.4.2 Normalized power spectrum

The noise power spectrum (NPS) was measured using the following method as described by Stierstorfer and Spahn (1999) and deVilliers and deJager (2003):

- Non overlapped sub-images of 128x128 pixels were obtained from the centre of images acquired at different exposures.
- Using a window of 128x128 pixels from the top-left-hand corner of the acquired image and a uniform grid, a two dimensional surface fit was produced and subtracted from each sub-image to obtain the low frequency background trends.
- Each column and row of a noise image was truncated by a scaled hanning filter of the form:

$$f(x) = 0.5[1 - \cos(2\pi x/N)] \quad (4.3)$$

$$0 \leq x \leq N$$

- The absolute magnitude squared of the two-dimensional Fast Fourier Transform was obtained for each sub-image.
- The one-dimensional NPS was obtained by averaging 15 rows around the axis on the two-dimensional image.
- The frequency values were transformed to the radial frequency by applying:

$$f_{radial} = \sqrt{f_x^2 + f_y^2} \quad (4.4)$$

Figure 4.12 shows the measured NPS for different detector exposures.

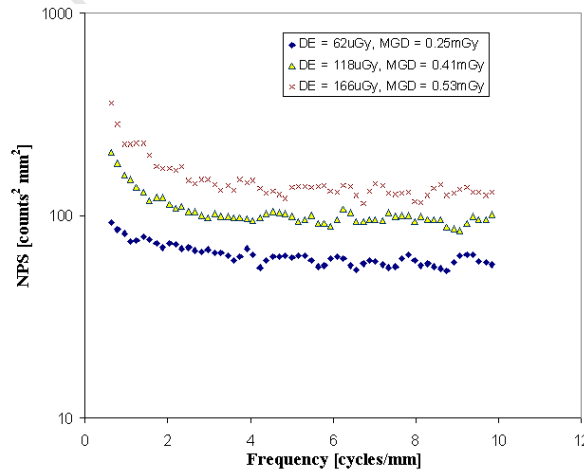


Figure 4.12: The measured noise power spectrum (NPS) for different detector exposures.

4.2.4.3 Detective quantum efficiency

The detective quantum efficiency was calculated using the measured MTF and NPS and the formula described by deVilliers and deJager (2003):

$$DQE(f) = G^2 MTF^2(f) \frac{Q}{NPS(f)} \quad (4.5)$$

G is the incremental gain, which is determined from the relationship between the mean intensity and the estimated photon flux on the detector; G is equal to the slope of the detector characteristic curve (shown in Figure 4.13); MTF is the measured system MTF; Q is incident x-ray photons per unit area on the detector, NPS is the measured noise power spectrum. Figure 4.14 shows the measured DQE for different detector exposures.

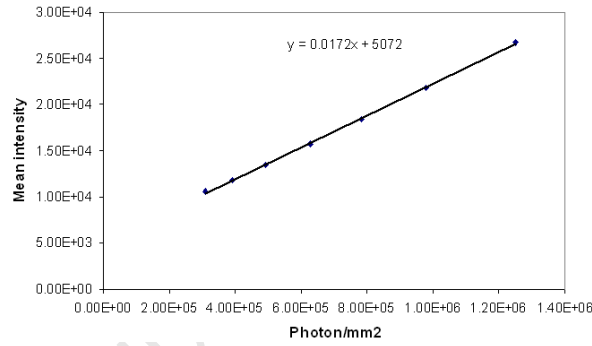


Figure 4.13: The detector characteristic curve used for calculation of the incremental gain.

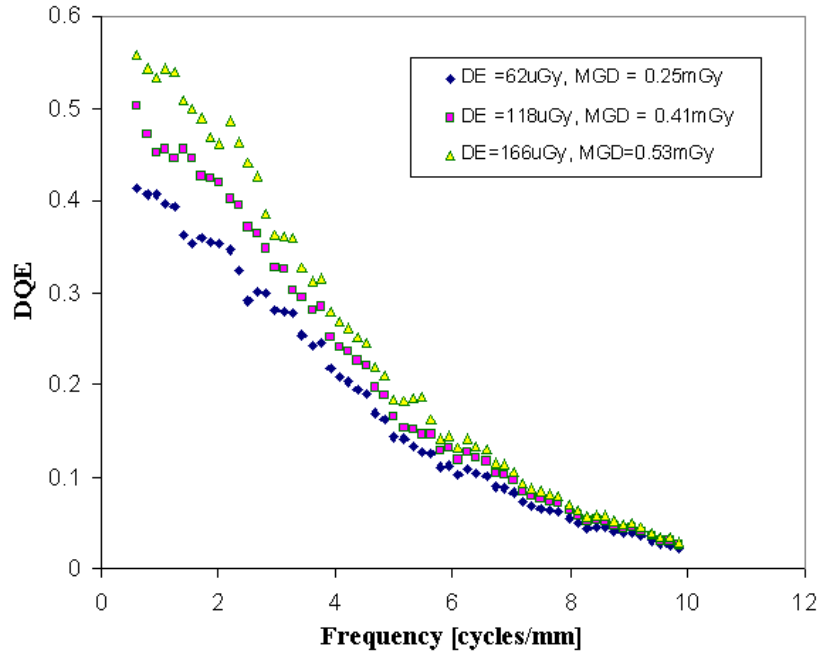


Figure 4.14: The detective quantum efficiency (DQE) for different detector exposures.

4.2.5 System evaluation using a breast tissue-equivalent phantom

The physical breast phantom was used to evaluate the LSSM system performance in terms of visual assessment, image contrast measurements, system spatial resolution, and the level of detector noise. The phantom used in the evaluation contains high contrast objects (group of microcalcifications), low contrast objects (hemispheric tumour masses), fibrous calcifications in ducts and masses simulating tumours, step wedge areas simulating the absorption of different glandular and adipose tissues, in addition to line pair tools. Figure 4.15 shows the layout of the breast phantom. The entrance dose was measured at the surface of the phantom using the Radcal ionisation chamber.

4.2.5.1 Visual assessment

Several phantom images were acquired using different mammography systems available in Cape Town including: (1) film-screen (GE Senographe 600T); (2) computed radiography CR (GE Performa); and (3) digital mammography (Siemens Novation) for comparison with our LSSM performance. All images were taken at the best technique

factors normally used in each department for the same breast thickness. The film from the film-screen mammography system was scanned using a high resolution scanner. Figure 4.16 shows phantom images taken from the different mammography systems. The system performance was evaluated by two chief medical physicists, an experienced radiographer and experienced radiologist. The observers were informed about the layout of the tissue-equivalent breast phantom used.

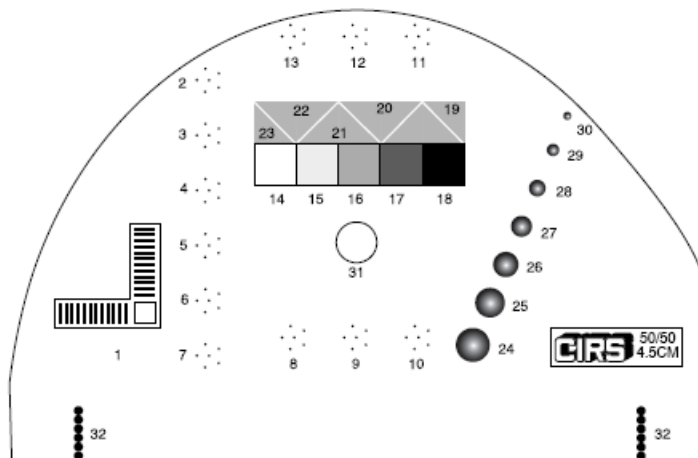


Figure 4.15: The tissue-equivalent phantom layout specifications taken from the CIRS data sheet for Model 011A, see appendix D.

4.2.5.2 Contrast measurement

Evaluation of system performance using the physical tissue-equivalent phantom shows the ability of the LSSM system to detect microcalcifications (high contrast objects) of 0.165mm and hemispheric masses (low contrast objects) of 1.59mm at mean glandular dose (MGD) of 0.3mGy with a good resolution of nearly 9lp/mm. The smallest microcalcifications of 0.130mm and hemispheric masses of 1.19mm can be detected at MGD of 0.5mGy. Figure 4.17 shows the detectable microcalcifications for each mammography system in images taken at technique factors described in the evaluation summary in Table 4.3. The smallest microcalcifications visible in the LSSM image at mean glandular dose of 0.3mGy, are 165 μ m in size.

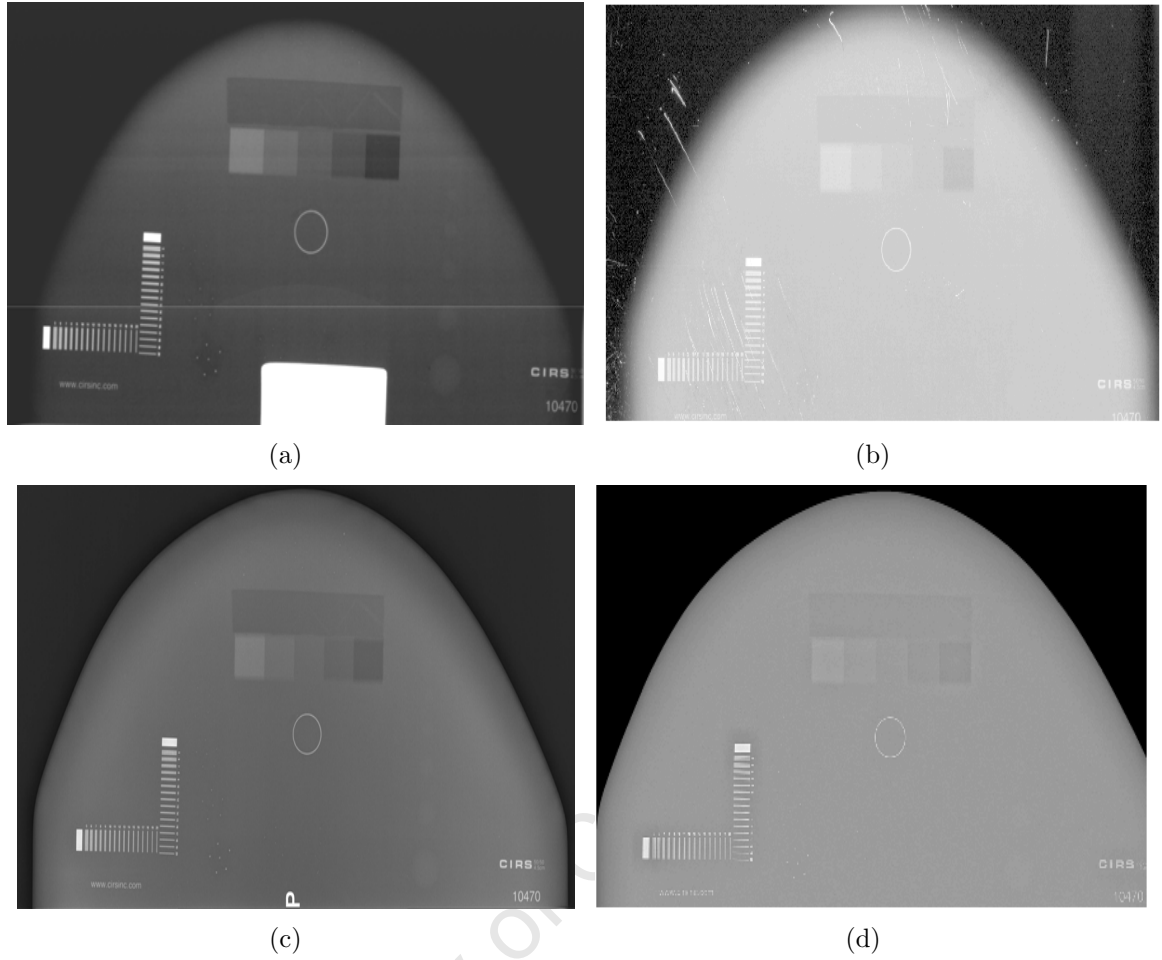


Figure 4.16: Phantom image from LSSM system compared to images from some existing mammography systems; (a) LSSM; (b) film-screen (GE Senographe 600T); (c) computed radiography (CR) (GE Performa); (d) digital mammography (Siemens Novation).

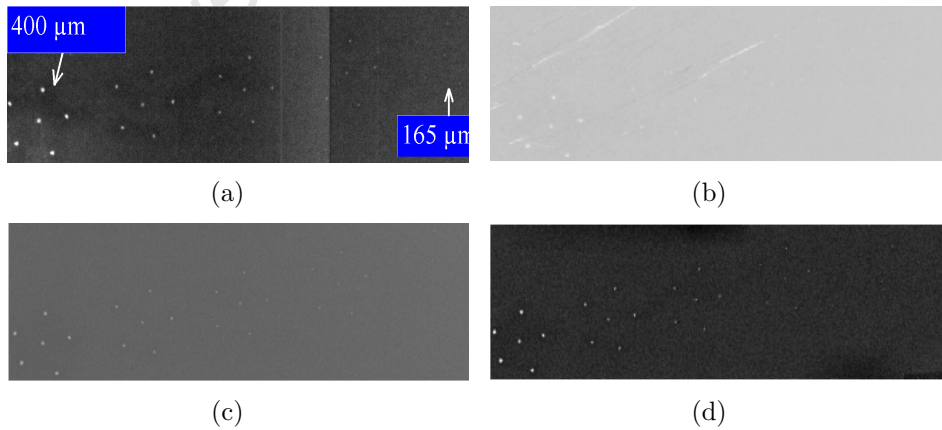


Figure 4.17: High contrast assessment using phantom images; (a) LSSM image; (b) analogue image; (c) CR image; (d) digital image; images were obtained using the technique factors shown in Table 4.3.

4.2.5.3 System resolution

The system resolution was evaluated visually by the observers mentioned in Section 4.2.5.1, using the line pair tool within the breast phantom and the Lodox Statscan line pair tool. Figure 4.18 shows the measured system resolution using the line pair tool in the breast phantom. The resolution of the analogue mammography system was more than 11lp/mm, but due to the blurring caused by scanning of the x-ray film the resolution was degraded to less than 5lp/mm as shown in Figure 4.18(b). The digital system shows a good contrast with lower resolution compared to other systems. The LSSM system shows a resolution of 9lp/mm. The resolution of the LSSM detector was also measured using the Lodox line pair tools and was found to be 9lp/mm in the slot direction and 8lp/mm in the scan direction as shown in Figure 4.19.

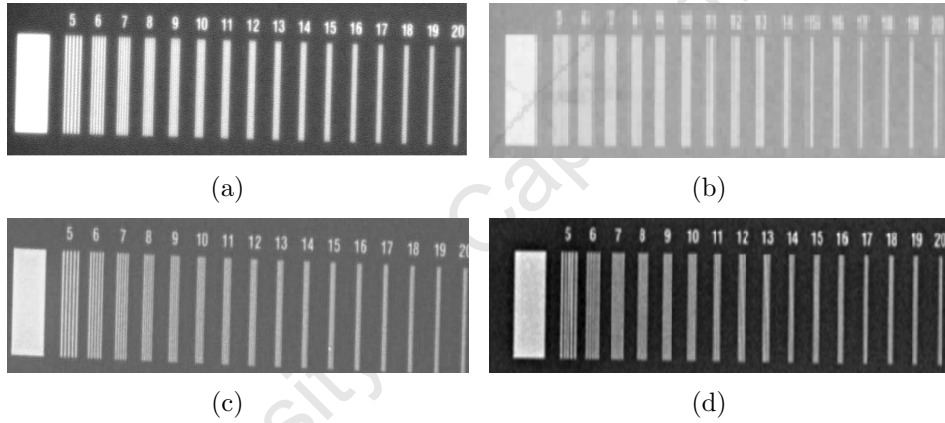


Figure 4.18: High contrast assessment using phantom images; (a) LSSM image; (b) analogue image; (c) CR image; (d) digital image.

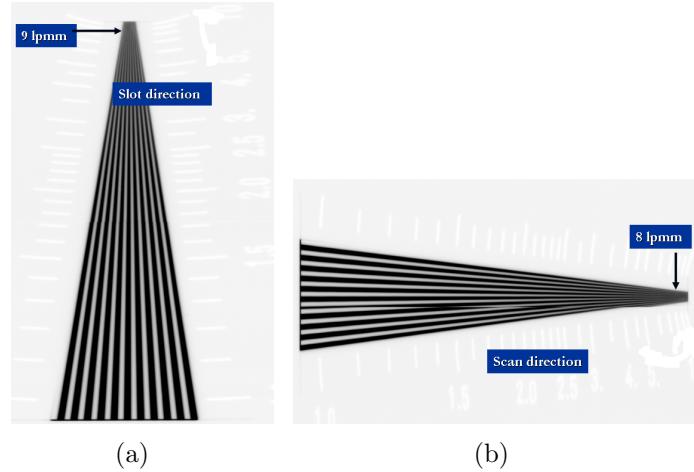


Figure 4.19: Detector resolution using the Lodox Statscan line pair tool; (a) slot direction; (b) scan direction.

4.2.5.4 Noise level

The detector noise is a very important parameter that affects the performance of the detector. High detector noise will result in a low contrast-to-noise ratio (CNR) and hence high dose to the patient and vice versa. The level of detector noise increases with increasing x-ray fluence. The relative noise level (the ratio of the detector noise to the mean signal corrected by subtracting the dark signal) decreases with increasing x-ray fluence. The higher the electronic dark signal, the higher the noise level. The additive noise is also a factor affecting the detector performance and hence the image quality of the system. Increasing the electronic noise decreases the image contrast and hence the ability of the system to detect a small microcalcification. Figure 4.20 shows the detector additive noise expressed in variance, for three detectors under evaluation: LSSM, Lodox Statscan, and Fischer slot scanning mammography (SenoScan). The additive noise for the Statscan and SenoScan were estimated from Scheelke (2005) and Tesic *et al.* (1999) respectively.

Further evaluation of the LSSM detector was done using a set of images of the breast phantom taken using Lodox Statscan and LSSM at the same technique factors (nearly equal tube voltage and the same entrance exposure). Figure 4.21 shows the CNR of 100% glandular tissue measured for LSSM and Lodox Statscan. Lodox Statscan shows good CNR compared to LSSM for the same dose. To achieve the same CNR, the noise level in the LSSM detector needs to be reduced to a level equal to that of Lodox; this can be achieved by increasing dose as shown in Figure 4.22.

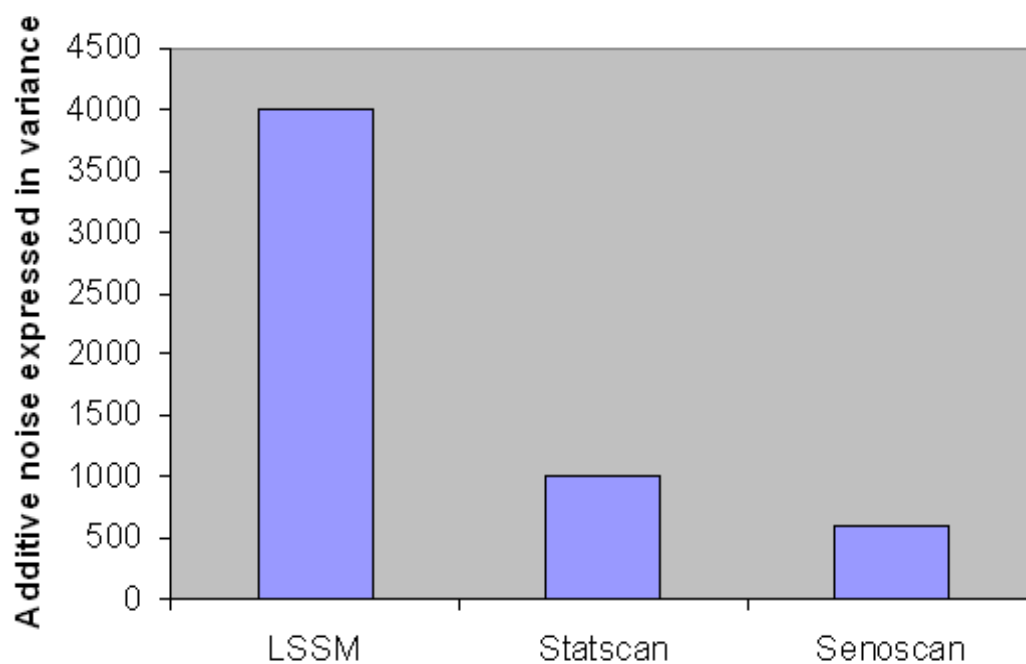


Figure 4.20: Detector additive noise for three different mammography systems.

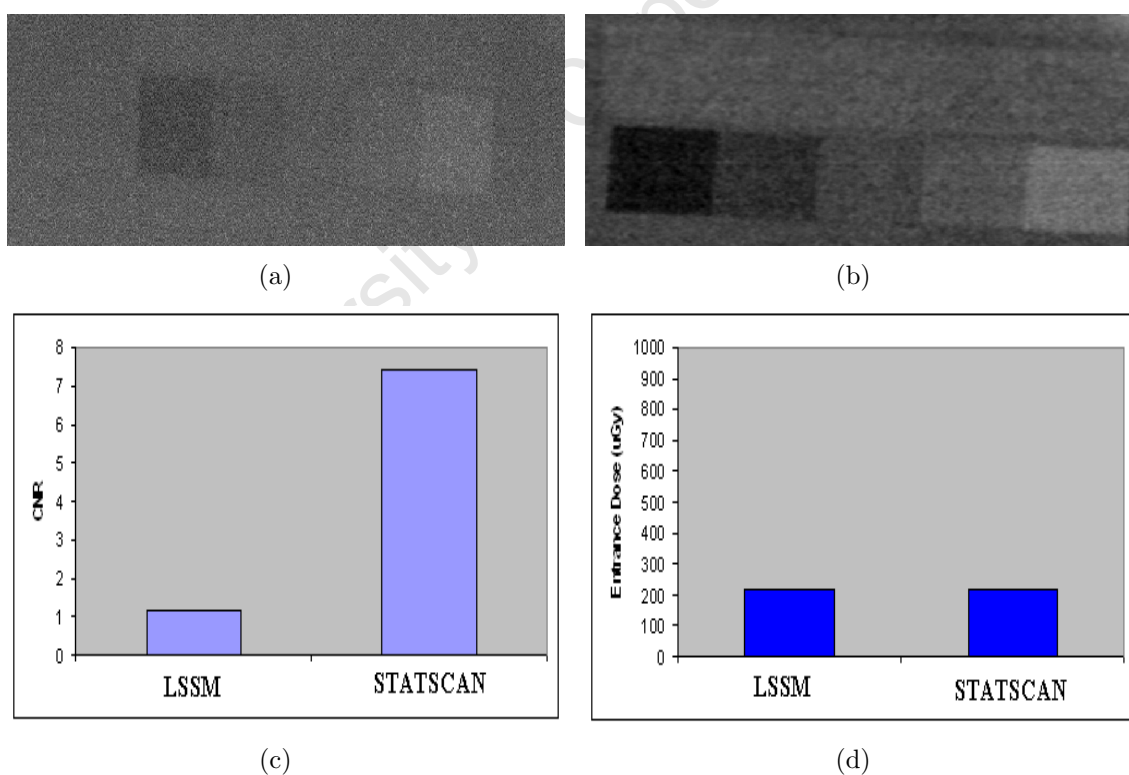


Figure 4.21: The effect of detector noise on the contrast-to-noise ratio (CNR) using phantom images of a step wedge taken at an entrance exposure of $200\mu\text{Gy}$; (a) LSSM image; (b) Lodox Statscan image; (c) CNR; (d) entrance exposure.

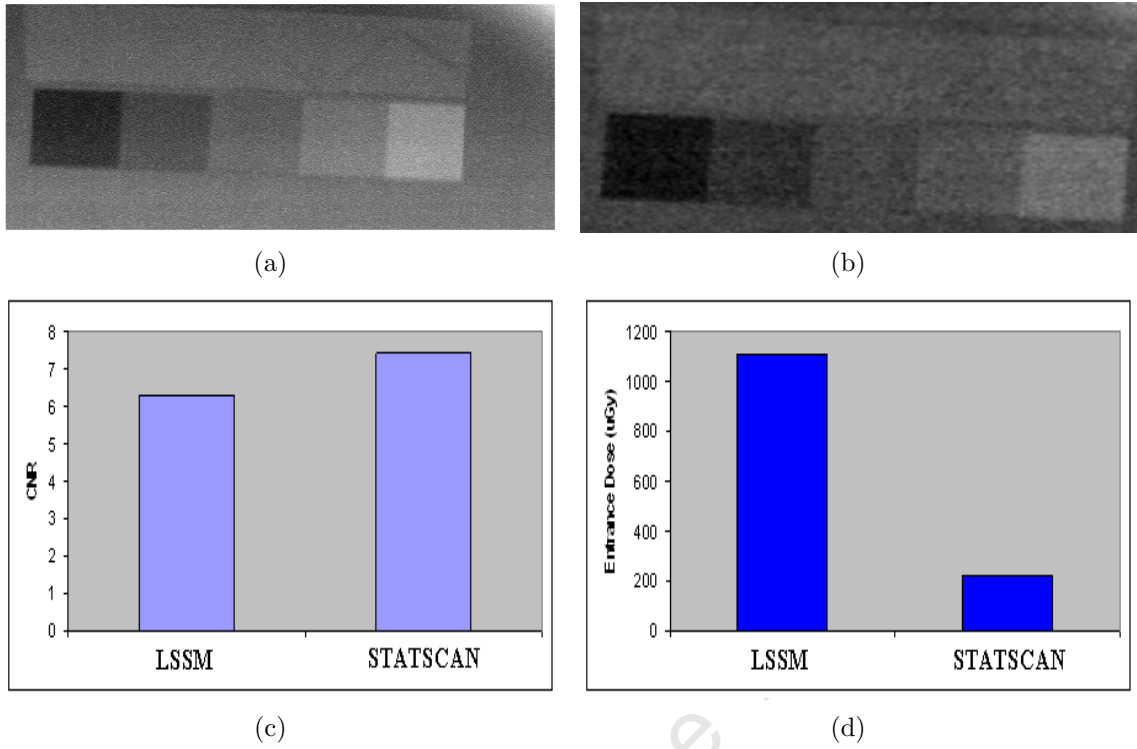


Figure 4.22: The effect of increasing entrance the dose for LSSM to $1100\mu\text{Gy}$, keeping the dose for the Lodox image the same; (a) LSSM image; (b) Lodox Statscan image; (c) contrast-to-noise ratio; (d) entrance exposure.

4.3 System Comparison Summary

Table 4.3 shows the summary of the evaluation of the slot scanning mammography system (LSSM) using the tissue-equivalent breast phantom. The performance of LSSM was compared to that of mammography systems currently in use in Cape Town, and of published data for the Fischer SenoScan slot scanning mammography system (Tescic *et al.*, 1999). The contrast, noise level, and the contrast-to-noise ratio (CNR) were calculated for 100% glandular tissue. The entrance dose was measured on the surface of the phantom.

Table 4.3: Summary of the parameters used for digital images obtained of the physical phantom; the figures in brackets represent tube current mA rather than exposure time mAs. The minimum CaCO_3 grain and hemispherical mass sizes visible at these technique factors are shown.

	Film-screen	CR	Digital	LSSM	Fischer
SDD[cm]	60	60	65	65	60
kVp	26	31	31	33	31
mAs (mA)	20	50	15	(100)	(200)
Entrance Dose [mGy]	1.5	3.93	1.34	1.33	6
Mean glandular Dose [mGy]	0.44	1.17	0.4	0.39	1.79
Resolution [lp/mm]	10	9	6	9	10
CaCO_3 grain size [mm]	0.196	0.165	0.165	0.165	
Hemi Masses [mm]	1.98	1.98	1.59	1.59	
Contrast		0.08	0.04	0.05	
Image Noise Level		0.018	0.011	0.013	0.0019
CNR		5.5	4.10	4.0	

4.4 Summary and discussion

The evaluation of the slot scanning digital mammography system (LSSM) using a tissue-equivalent phantom designed for mammography shows good performance compared to the existing systems examined. The LSSM detector shows a good linearity response of 99.94% to the entrance exposure over the range 0.3 to 3.0mGy. The visual evaluation using the CaCO_3 grain and hemispherical masses within the tissue-equivalent breast phantom show that the LSSM approaches the resolution of analogue mammography, with much better contrast. The LSSM shows an equivalent contrast and resolution at half the dose of computed radiography (CR). LSSM shows good contrast and much better resolution at an equivalent dose to that of the Siemens Novation digital mammography system. The LSSM system shows a detective quantum efficiency (DQE) of 0.56 at mean glandular dose of 0.5mGy, which is comparable to that of existing digital mammography systems that use the indirect detection method, but is less than that of systems based on direct and photon counting detectors. Although the LSSM shows good performance compared to the existing mammography systems, the evaluation of detector noise shows a high level of noise when compared to linear slot scanning radiography (Lodox Statscan). The estimated electronic noise for LSSM was found 4850e compared to the 1000e for the Lodox Statscan (Scheelke, 2005) and less than 2 DU (using the Fischer detector gain, this value is approximately equivalent to 600e) (Tesci *et al.*, 1999) for Fischer slot scanning mammography (SenoScan). The LSSM system

show a noisier detector compared with other systems, which may be due to the fact that the x-ray detector of the LSSM system was not especially optimized for mammography applications. Therefore further optimization of the detector in terms of noise will result in much better contrast with even lower radiation dose.

4.5 Author's contribution

Using the system requirements and design specifications derived by the author, the author and the mammography team in the MRC/UCT Medical Imaging Research Unit, University of Cape Town, developed a prototype mammography system (LSSM) using the linear slot scanning technique. The author performed the system comparison described in this chapter.

Chapter 5

Modelling and Validation of a Mathematical Tissue-Equivalent Breast Phantom for Digital Mammography

5.1 Introduction

Quantitative image quality parameters such as modulation transfer function (MTF), noise power spectrum (NPS), detective quantum efficiency (DQE), signal to noise ratio (SNR), and figure of merit (FOM) are well known in the assessment of digital mammography detector performance (Cunningham, 1994; Siewerdsen and Antonuk, 1998; Siewerdsen *et al.*, 1998; Stierstorfer and Spahn, 1999; Siewerdsen and Jaffray, 2000; Tkaczyk *et al.*, 2001; Cunningham *et al.*, 2002; Neitzel *et al.*, 2004). Several test phantoms have been developed for performance assessment of digital mammography, such as the ACR (Barnes and Hendrick, 1994), TRO (MAM)(Cowen and Coleman, 1990), and IQI (Hessler *et al.*, 1985) test phantoms. Images produced by these test phantoms are subjectively assessed by a human observer. The differences among human observers in the visibility thresholds of structures such as microcalcifications result in variations in perceived system performance. This limitation gives rise to a need for automatic objective assessment. Several objective methods that have been developed to quantify

system performance using computational software to emulate human detection, and are comparable with but more consistent than human observers (Chakraborty and Eckert, 1995; Smith *et al.*, 1998; Verdun *et al.*, 1996; Veldkamp and Karssemeijer, 1994; Pachoud *et al.*, 2004). Physical image acquisition is required, and these objective methods mainly depend on the sensitivity of the physical phantom to x-ray exposure, which will affect the quantitative assessment of image quality. These limitations indicate a need for a mathematical phantom with which to simulate phantom images for system design and system optimization. Several 2D and 3D mathematical simulations of breast tissues and microcalcifications have been developed (Bakic *et al.*, 2002; Bliznakova *et al.*, 2003; Hunt *et al.*, 2005; Dance *et al.*, 2005; Reiser *et al.*, 2006). Bakic *et al.* (2002) presented a model to simulate 2D mammograms based on simulated 3D anatomy. Hunt *et al.* (2005) and Dance *et al.* (2005) used a voxelised phantom to estimate mean glandular dose for breast dosimetry. Bliznakova *et al.* (2003) simulated a 3D non-compressed breast containing tissues of different size, shape and composition, in order to generate images resembling real mammograms. They assumed the only source of noise to be the natural fluctuation of the photon flux (quantum mottle).

This chapter presents a tissue-equivalent mathematical phantom for digital mammography. Some structures in the mathematical phantom were simulated similarly to those in the physical tissue-equivalent phantom designed for mammography (Model 011A) manufactured by CIRS (Computerized Imaging Reference Systems, Norfolk Virginia 23513, USA) available in the MRC/UCT Medical Imaging Research Unit. All measurements were taken from the recently developed linear slot scanning mammography system (LSSM). All programs were coded by the author using Matlab (Matlab, The MathWorks, Natick, MA).

5.2 Methods

A successful x-ray phantom should mimic the attenuation properties of the tissues it represents. The mathematical breast phantom was simulated using breast tissue equivalent material of 50:50 glandular/adipose tissue. The visibility of structures within the phantom is represented by their contrast against the breast tissue. As shown in Figure 5.1 and Table 5.1, five areas containing structures simulating the absorption of 100% glandular, 70:30 glandular/adipose, 50:50 glandular/adipose, 30:70 glandular/adipose, and

100% adipose tissues, four hemispheric regions and specks simulating tumour masses and microcalcifications of different sizes are included in the phantom. The intensities of structures were calculated from the generated target signal from the attenuated beam through the breast tissue and the structures, assuming constant attenuation in the pixels covering the area of each object. The partial volume effect was ignored in this simulation due to the fact that the scanning technique can cover the whole area of simulated target tissue. The mathematical phantom image was modeled using the following steps:

- Modelling the pixel intensity
- Estimating the object contrast
- Modelling the detector noise
- Modelling the total system resolution

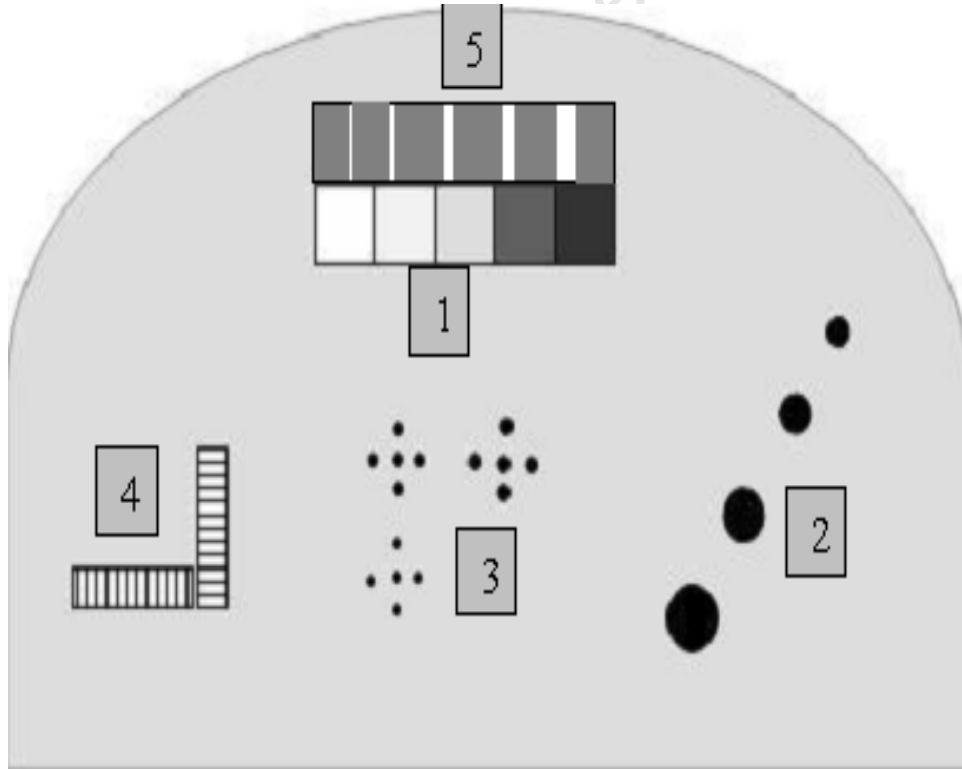


Figure 5.1: The mathematical tissue-equivalent phantom for digital mammography. Areas within the phantom simulating (1) 100% glandular, 70:30 glandular/adipose, 50:50 glandular/adipose, 30:70 glandular/adipose, and 100% adipose tissues, from left to right. (2) four hemispheric regions; (3) microcalcifications of different sizes ($CaCO_3$); (4) line pair tools for measuring the system resolution; (5) nylon fibres of different size.

Table 5.1: Mathematical phantom specifications

Structures	Specifications
(1) Step wedge	100% glandular 70% glandular 50% glandular 30% glandular 100% adipose
(2) Hemispheric masses 75% glandular/25% adipose, thickness	1mm 2mm 3mm 4mm
(3) Calcification($CaCO_3$) grain size	0.122mm 0.170mm 0.400mm
(4) Line pair tools	8 lp/mm
(5) Nylon fibre	1.2mm 0.8mm 0.6mm 0.4mm 0.3mm

5.2.1 Modelling the pixel intensity in digital units

The pixel intensity T_e of the mathematical breast phantom in digital units (ADU) is proportional to the photoelectron intensity (S_e) generated by the x-ray photons incident on the detector and the added signal (dark signal) due to thermal electrons generated in the charge-coupled device (CCD) (D_s):

$$T_e \propto (D_s + S_e) \quad (5.1)$$

The signal (photoelectrons, S_e) generated by the x-ray detector from the attenuated x-ray photons depends on the following:

- (i) The efficiency of the detector scintillator, characterized by its quantum efficiency (QE_s).
- (ii) The conversion of x-ray energy into light photons, characterized by the scintillator gain ($DetGain$).

(iii) The efficiency of transfer of quanta to the CCD, characterized by the detector coupling efficiency (CE).

(iv) The integration of quanta by the CCD, characterized by the active pixel area of the CCD (A_P).

Using linear cascaded system analysis and the propagation of a signal through the cascaded stages of the imaging system (Cunningham, 1994; Maidment *et al.*, 1993; Siewerdsen and Antonuk, 1998; Siewerdsen *et al.*, 1998; Neitzel *et al.*, 2004), the photoelectrons S_e collected by the CCD pixel can be estimated from the calibrated photon flux incident on the detector and the total system gain.

$$S_e = QE_s \times CE \times DetGain \times QE_{CCD} \times A_P \int_{E_{min}}^{E_{max}} q_{Slot}(E) e^{-\mu_B x} dE \quad (5.2)$$

where $q_{Slot}(E)$ is modified energy spectrum for a slot scanning beam incident on the detector face and is given by Equation 3.1; μ_B is the energy-dependent total linear attenuation coefficient of breast tissue; QE_{CCD} is the CCD quantum efficiency; E_{min} and E_{max} are the minimum and the maximum photon energies in the x-ray spectrum.

Finally, the pixel intensity in digital units (ADU) can be estimated from the total detector signal and the analogue-to-digital conversion gain $ADCGain$:

$$Pix_{Int}(ADU) = \frac{S_e + D_s}{ADCGain} \quad (5.3)$$

$$Pix_{Int} = \frac{1}{ADCGain} S_e + S_D \quad (5.4)$$

S_D is the added signal in ADU estimated from the CCD and ADC data sheets or measured using minimum exposure technique settings.

5.2.2 Estimation of the image contrast

The contrast of an object in a mammography image depends on the density of the object, the total attenuation by the object and the soft tissue in the breast, the x-ray spectrum (photon energy and tube voltage kVp), scattered radiation, and the display

conditions. The object contrast can be estimated from the difference in the attenuation coefficients of the object and the surrounding background using Dobbins's relationship (Dobbins, 2000):

$$C = \frac{1}{1 + SPR} \left(1 - \frac{\int q_{Slot}(E) e^{-\mu_B(D-d) - \mu_A d} \eta(E) dE}{\int q_{Slot}(E) e^{-\mu_B D} \eta(E) dE} \right) \quad (5.5)$$

where $q_{Slot}(E)$ is the photon fluence of the slot beam; $\eta(E)$ is the energy absorption of scintillator; SPR is the scatter-to-primary ratio, which is disregarded here, as slot scanning eliminates more than 95% of the scattered photons; μ_A is the energy-dependent linear attenuation of target tissue; μ_B is the energy-dependent linear attenuation of background tissue; D is the background attenuated length; and d is the target attenuated length.

The image contrast of a digital system is defined as the relative difference in intensity between the target tissue (e.g. 100% glandular tissue) and the surrounding region (e.g. 50:50 glandular/adipose tissue) as shown in Equation 5.6. In a digital system the image contrast is equivalent to the object contrast, while it is modified by the film gradient in a screen-film system.

$$C_{img} = \frac{S_{target} - S_{background}}{S_{background}} \quad (5.6)$$

S_{target} is the mean intensity value in the target area; $S_{background}$ is mean intensity value in the surrounding background.

5.2.3 Modelling the detector noise

Due to the natural fluctuation of the incident photon flux, the electrons generated by the detector obey Poisson statistics. The self-generated noise (quantum noise) and the added noise can then be estimated by using the Poisson and Gaussian distributions. The pixel intensity of a generated image must correspond to the number of incident photons per pixel and the randomly distributed noise. A noisy phantom image can be modelled as follows, with the standard deviation (σ) representing the total amount of detector noise, while $n(x,y)$ is a matrix of x rows and y columns of random numbers corresponding to image size:

$$Ph_n = M_{val} + \sigma n(x, y) \quad (5.7)$$

Ph_n is the noisy phantom image, while M_{val} is the mean intensity pixel value corresponding to the electrons generated by the detector. The standard deviation represents the fluctuation of the pixel intensity due to the quantum and dark noise. The quantum noise was estimated by the square root of the generated detector signal. The dark noise can be estimated from the measurement of an image without x-rays (shutter closed) or from a sequence of images of added aluminium sheets using the minimum exposure technique settings available in the machine.

5.2.3.1 Sources of detector noise

There are several sources of noise that affect the contrast of the phantom image. The detector noise can be estimated from the following:

- Quantum noise (σ_Q) due to the natural fluctuation of the photon flux can be estimated by the square root of the the generated detector signal:

$$\sigma_Q = \sqrt{S_e} \quad (5.8)$$

- Additive electronic noise is all other electronic noise including: noise associated with dark current in the CCD (σ_D); noise from the amplifier σ_{amp} ; digitization noise from the analog-to-digital converters (ADCs) (σ_{Dig}).
- The total detector noise can then be expressed in variance:

$$\sigma_T^2 = \sigma_Q^2 + \sigma_D^2 + \sigma_{amp}^2 + \sigma_{Dig}^2 \quad (5.9)$$

5.2.4 Modelling the total system resolution

The amount of blurring or degradation of the phantom image depends on the distortion operator, known as the point spread function (PSF), which is convolved with an unblurred image to provide a blurred image. The modulation transfer function (MTF) is the frequency domain of the spatial domain point spread function. The line spread function (LSF) can be estimated if the mathematical equation of the PSF is known (Ewert and Zscherpel, 2008). Ewert and Zscherpel (2008) presented a mathematical

conversion a PSF to LSF based on the literature (Dainty and Shaw, 1974). The LSF of a PSF of Gaussian shape is a Gaussian shape (Ewert and Zscherpel, 2008):

$$f(r) = a^2 e^{-\pi r^2 a^2}$$

$$\Rightarrow$$

$$f(x) = a e^{-\pi x^2 a^2} \quad (5.10)$$

The total distortion of x-ray imaging detectors is characterized by the MTF of the system, which is the distortion in various parts of the detector such as the focal spot MTF, the scintillator MTF, pixel blurring, and the blurring due to mismatch between the scanning velocity and the transfer rate of charge in the CCD.

Figure 5.2 shows the Fourier Transforms of the applied distortion operators (a Gaussian PSF), which represent the MTF of the imaging system, and their effect. The line pair tools are used to visualize the amount of phantom image blurring in response to the application of different distortion operators.

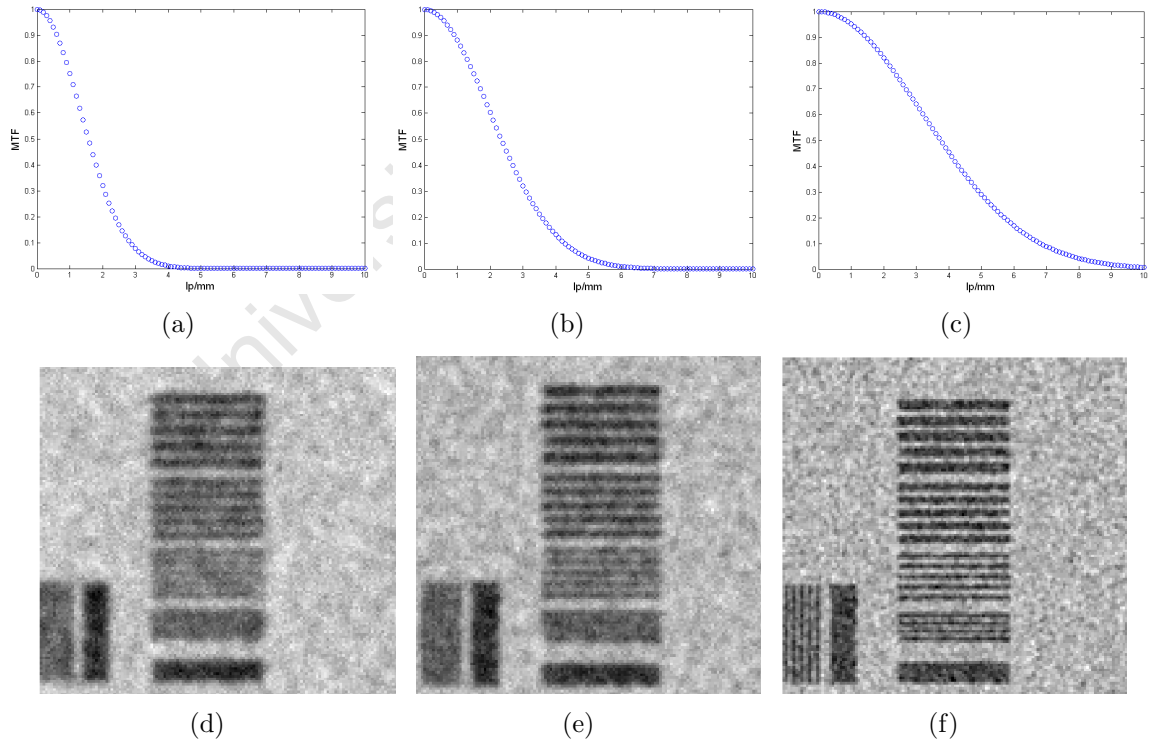


Figure 5.2: The effect of phantom image blur simulated for different distortion operators (Gaussian filters) on the line pair tools included in the mathematical phantom; (a), (b), and (c) are the Fourier Transforms of the applied distortion operators; (d), (e), and (f) indicate the amount of blurring for each operator.

5.3 Generation of phantom image

A grayscale intensity image is generated which contains the body of the phantom (50:50 glandular/adipose tissue) and areas representing the structures contained in the phantom as described earlier. The intensity of pixels within the phantom body or in each area is defined by the attenuation of each structure (as a function of its density, the linear attenuation coefficient, and the thickness) or by its contrast against the background. The phantom image parameters such as pixel intensity value, the location of structures within the phantom image, and the size and dimensions of the phantom are stored in a phantom parameter matrix. The phantom image can be modelled as follows:

- An initial phantom image is generated consisting of x rows and y columns with a default pixel intensity value equal to the intensity value corresponding to the digital value of the 50:50 glandular/adipose tissue signal.
- The pixel intensity value in digital units (ADU) is then modified for each structure according to its contrast.

$$Ph(x, y) = Ph_{in}(x, y) + A \quad (5.11)$$

$Ph_{in}(x, y)$ is the initial pixel intensity value; A is the intensity value calculated from the attenuation of each structure within the phantom.

- Quantum noise due to the nature of the x-rays is then added to the generated phantom image using Equation 5.7.
- The phantom image is then modulated using the system resolution (MTF) as a distortion factor.

Figure 5.3 shows a phantom image simulated using tube voltage of 30kVp, tube current of 200mA, and 4000 electrons of additive noise.

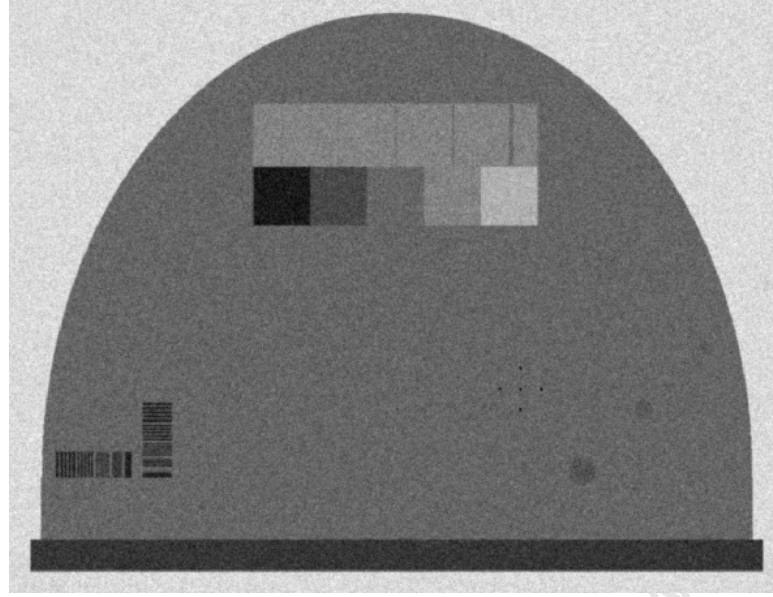


Figure 5.3: A phantom image generated using 30kVp, 200mA.

5.4 Validation of the mathematical phantom

The mathematical breast phantom was validated using a series of digital images of the CIRS physical tissue-equivalent phantom on which it was based. Phantom images were acquired using the prototype slot scanning mammography system (LSSM). The physical phantom was used to measure the pixel intensity, subject contrast, contrast-to-noise ratio (CNR), analogue-to-digital convertor (ADC) conversion gain, and detector additive noise. These measured values were used to validate the simulation of mathematical phantom images and the estimation of the detector noise and the conversion gain.

5.4.1 Estimation of digital pixel intensity and ADC conversion gain

The number of photoelectrons per pixel generated on the detector was estimated for different technique factors (tube voltage: 25, 30, 35 and 40 kVp; tube current: 16 - 200mA) using Equation 5.2. A series of digital images of the physical phantom were acquired at different technique factors as shown in Table 5.2. The mean pixel values in 50:50 glandular/adipose tissue in the phantom was measured for each digital image in a region of interest comprising a 100x100 pixel area. The pixel values for a tube

voltage of 30kVp at different tube currents were plotted against the number of estimated photoelectrons for each detector exposure (different mA) as shown in Figure 5.4.

Table 5.2: Parameters used in obtaining digital images of the physical phantom.

	parameters
Tube voltage	25, 30, 35, and 40 kVp
Tube current	16-200 mA
Scanning speed	22.5 mm/s
CCD Pixel binning	1x1
Source-to-phantom distance	60 cm
Collimator gap	0.8 mm
Pixel size	0.048 mm

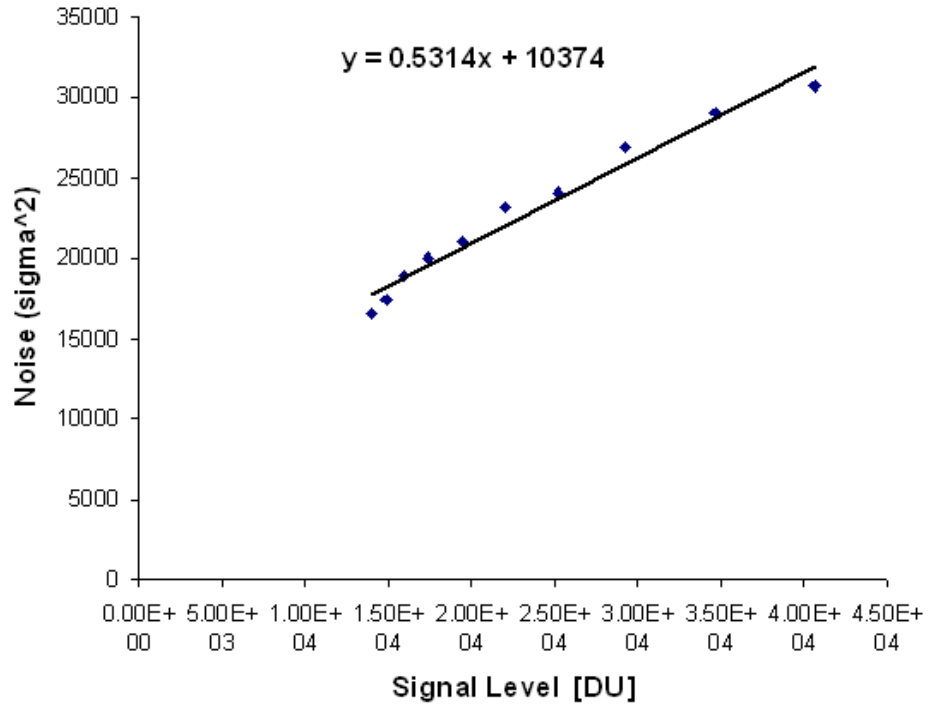


Figure 5.4: Least squares fit of the measured pixel intensity (signal level) and the square of standard deviation (noise level).

Using a least squares fit of the measured pixel intensities (in ADU) and the estimated detector signal in (e^-), the digital pixel values at different exposures can be estimated using the following equation:

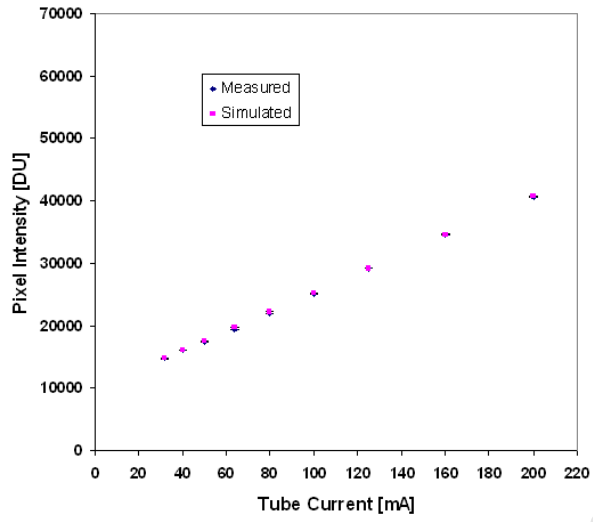
$$Pix_{intensity}(ADU) = 2.0298 \times Pix_{intensity}(e^-) + 9851 \quad (5.12)$$

The sensitivity of the detector is given by the slope ($2.0298ADU/e^-$) of the fitted line in Figure 5.4, which is slightly less than the value ($2.0e^-/ADU$) set manually by the operator.

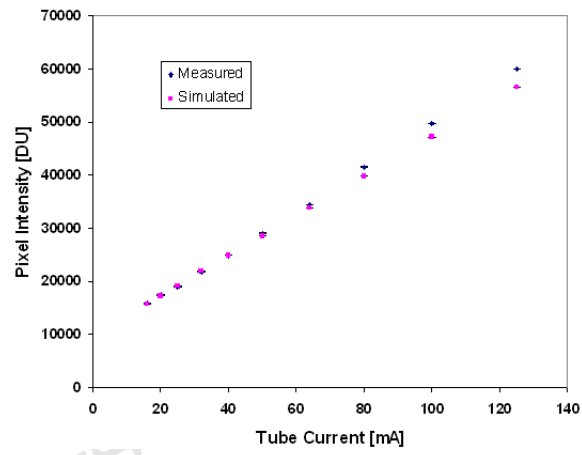
The conversion gain is given by the reciprocal of the slope and was found to be approximately equal to $0.5e^-/ADU$.

The additive detector signal (dark signal) can also be estimated for the fitted line and was found to be 9851 digital units (ADU) which is approximately equivalent to 4850 electrons. This value agreed (within 95%) with the measured pixel intensity for an image taken without x-ray exposure.

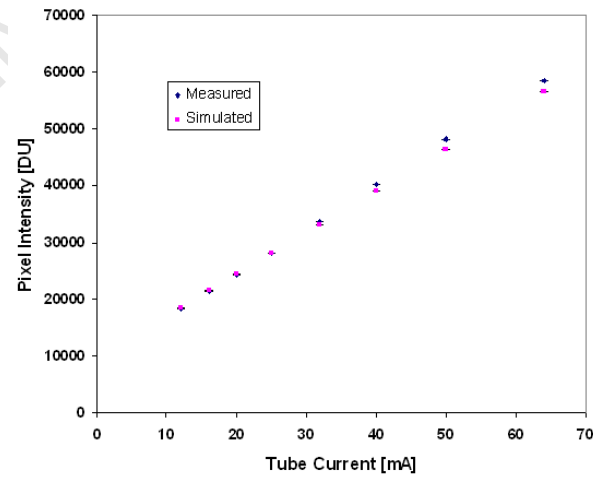
Figure 5.5 shows the pixel intensity estimated, using Equation 5.12, for tube voltages 30, 35 and 40 kVp, compared to the measured values. The error bars in Figure 5.5 indicate the range of the measured pixel intensity of the selected region of interest (i.e. the pixel intensity for each selected region is averaged over three readings).



(a)



(b)



(c)

Figure 5.5: The mean estimated and measured digital pixel intensity at (a) 30kVp (b) 35kVp (c) 40kVp.

5.4.2 Estimation of the image contrast

Figure 5.6 shows the estimated and the measured contrast for different structures in the phantom as a function of tube voltage (kVp); estimates were obtained using Equations 5.5 and 5.6.

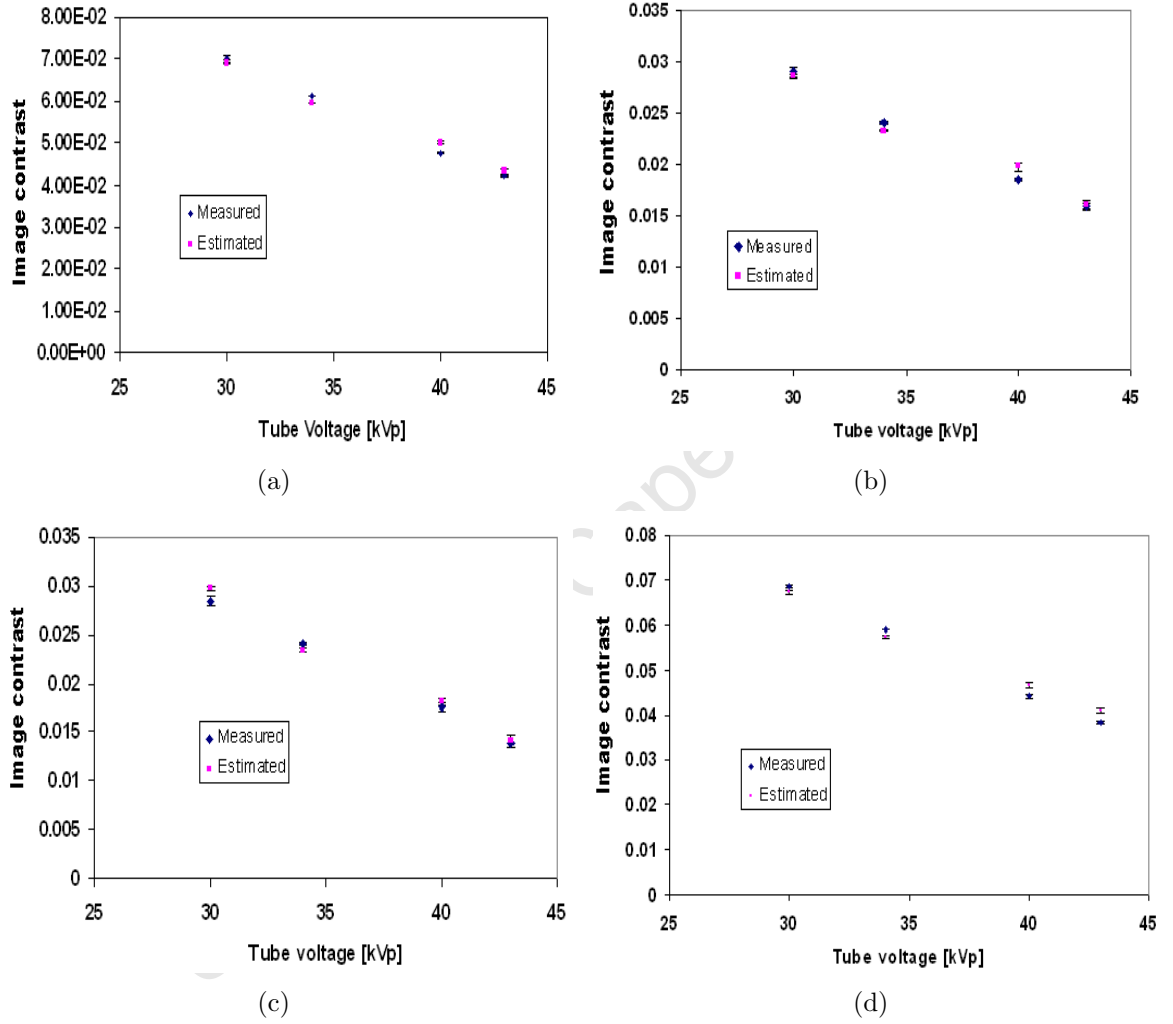


Figure 5.6: The estimated and measured image contrast of different structures in the phantom as a function of tube voltage (a) 100% glandular tissue (b) 30:70 glandular/adipose (c) 70:30 glandular/adipose (d) 100% adipose tissue. The error bars indicate the range in the image contrast, as three image contrast readings were taken at each setting.

Figure 5.7 shows the estimated and the measured noise as standard deviation for 50% glandular tissue in the phantom as a function of tube current.

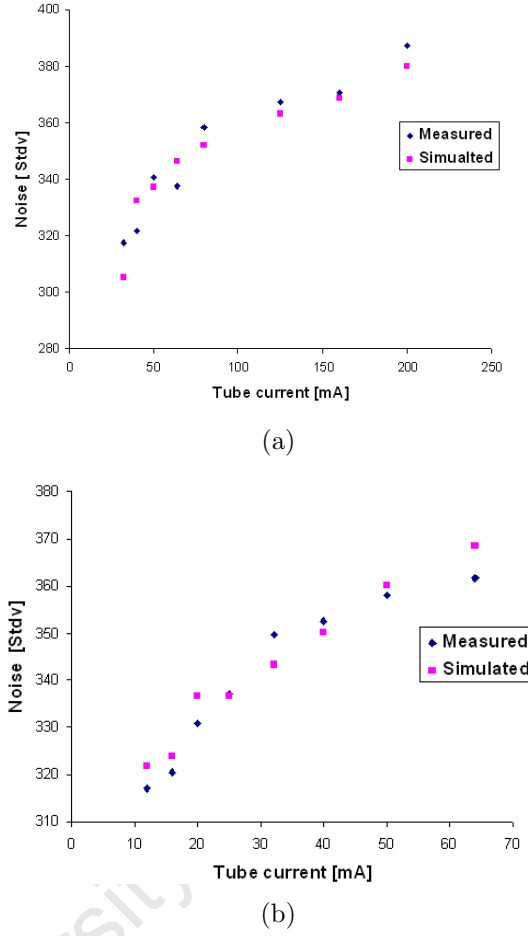


Figure 5.7: The estimated and measured image noise of 50% glandular tissue at (a) 30kVp (b) 40kVp.

5.4.3 Estimation of contrast-to-noise ratio

The contrast-to-noise ratio is the relative difference in signal between a target and the surrounding background divided by the inherent noise in the image (Equation 5.13). Noise usually is the measured standard deviation σ of a region of interest in the background.

$$C_{img} = \frac{S_{target} - S_{background}}{noise} = \frac{S_{target} - S_{background}}{\sigma} \quad (5.13)$$

Figure 5.8 shows a visual comparison between the estimated and the measured image contrast for different tissue-equivalent regions included in the mathematical phantom at different tube voltages and tube currents. Figures 5.9 and 5.10 show the simulated

contrast-to-noise ratio (CNR) for tube voltages 30 and 40 kVp compared to the measured CNR from the digital images of the physical phantom. The error bars in Figures 5.9 and 5.10 indicate the range for individually measured CNR in each region of interest (three readings were taken). Figure 5.11 shows the difference between the measured and simulated CNR as a percentage of their mean.

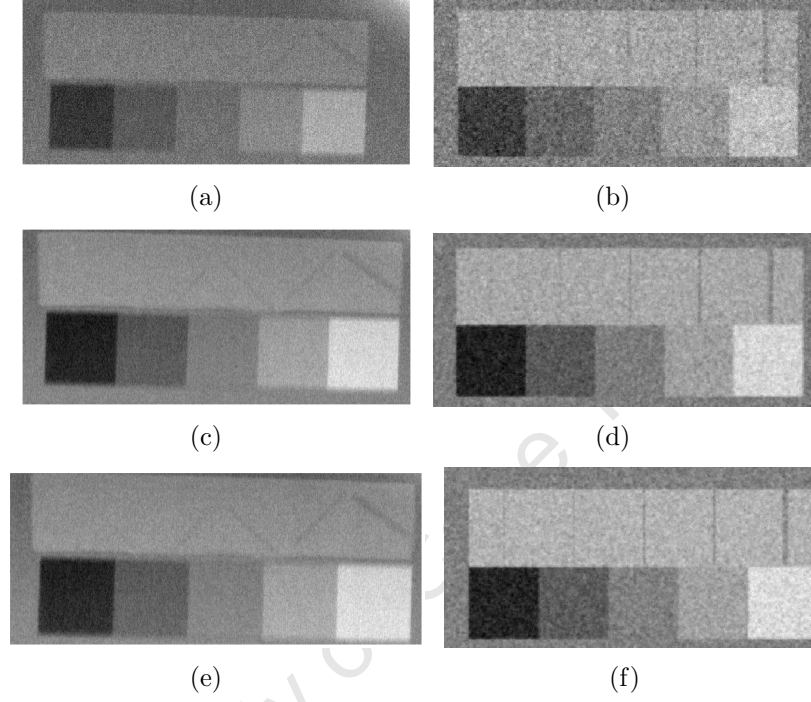


Figure 5.8: The mathematical and the physical phantom images: (a) measured 30kVp, 80mA; (b) simulated 30kVp, 80mA; (c) measured 35kVp, 100mA; (d) simulated 35kVp, 100mA; (e) measured 40kVp, 64mA; (f) simulated 40kVp, 64mA. The nylon fibres in the mathematical phantom were simulated having the same diameters as those in the physical phantom, but in a vertical rather than diagonal direction for ease of implementation.

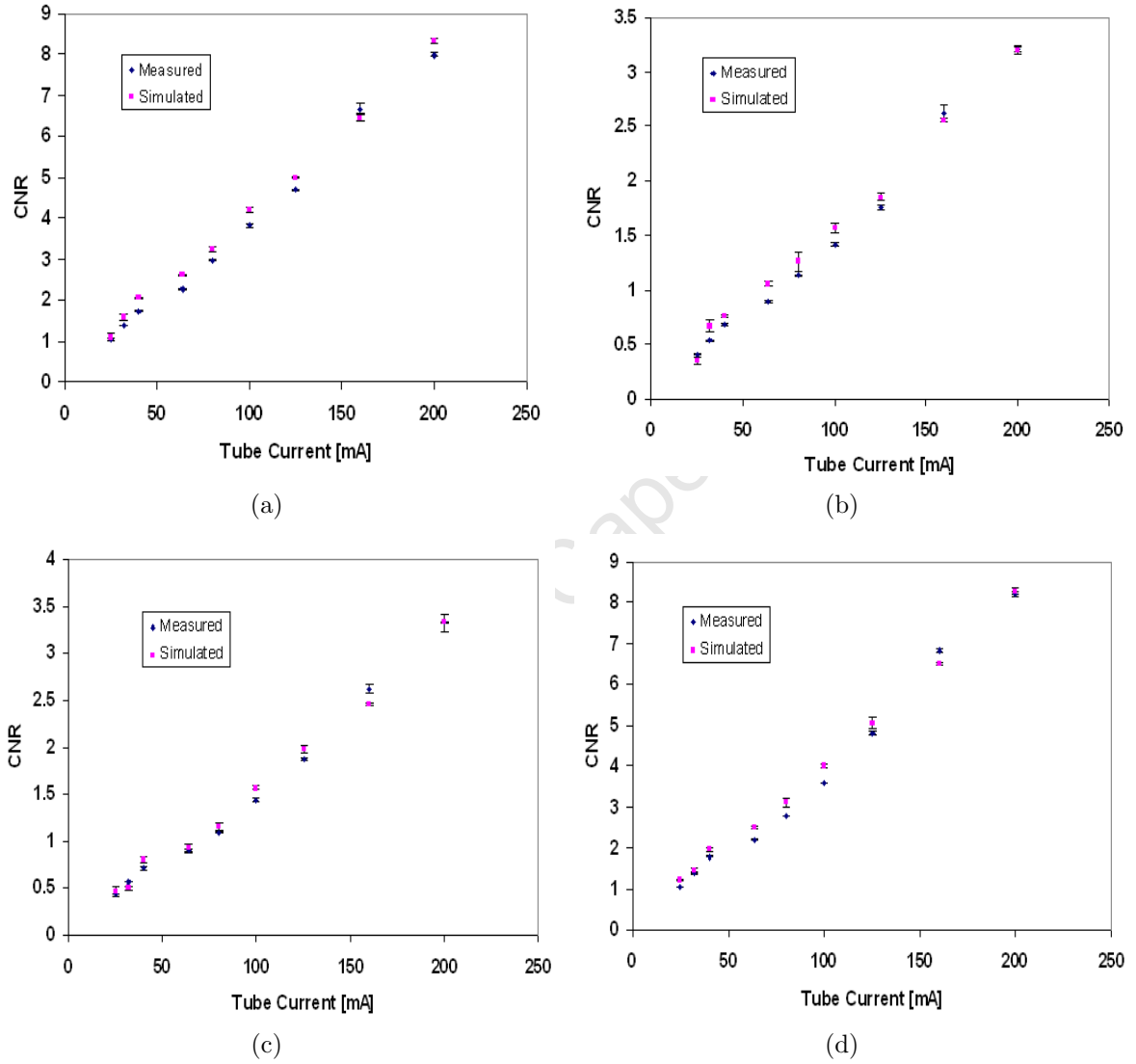


Figure 5.9: Measured and simulated contrast-to-noise ratio (CNR) for different tissues within the phantom image at 30kVp tube voltage; (a) 100% glandular tissue; (b) 70% glandular tissue; (c) 30% glandular tissue; (d) 100% adipose tissue. The error bars indicate the range in CNR in each region of interest.

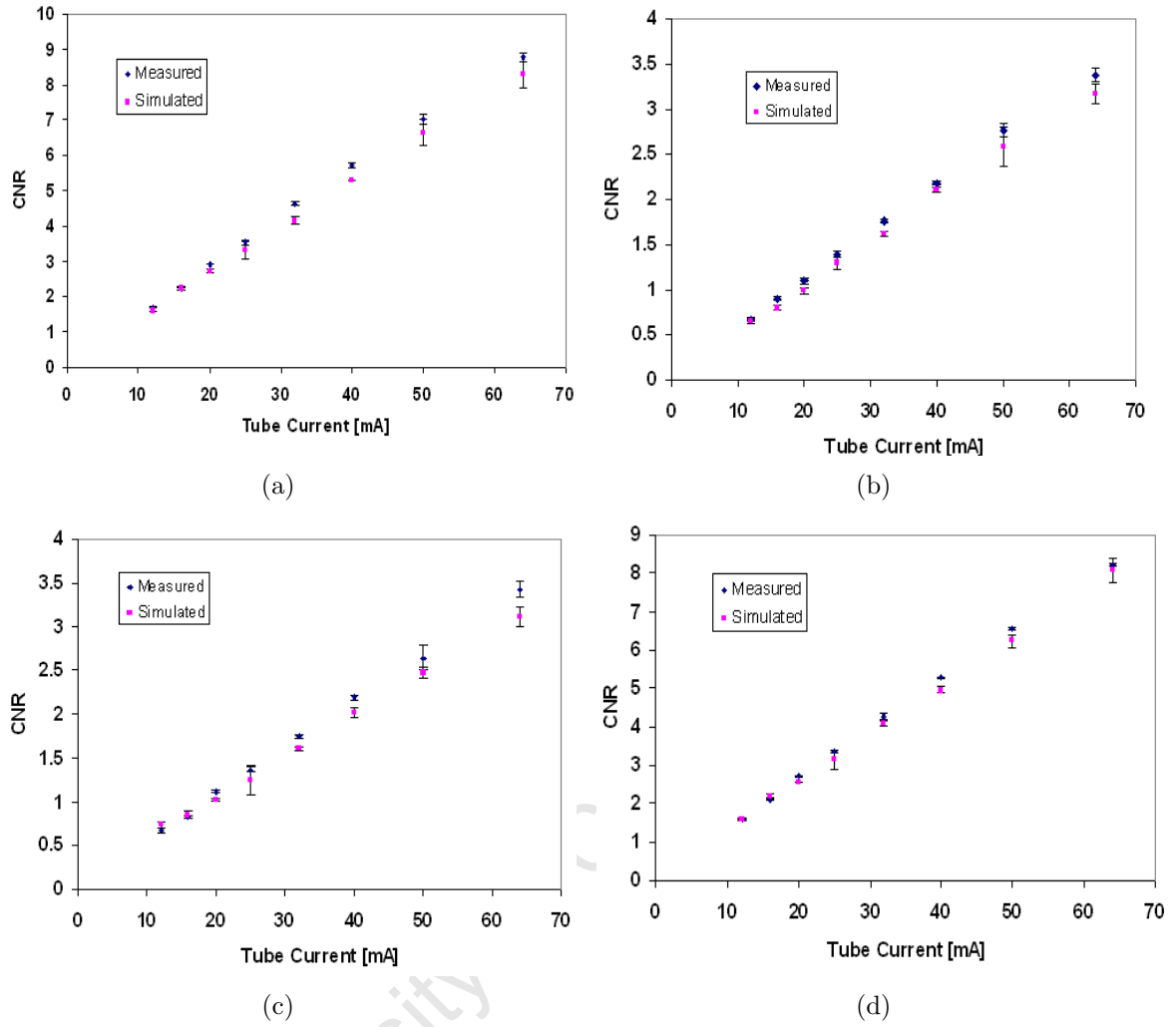


Figure 5.10: Measured and simulated contrast-to-noise ratio (CNR) for different tissues in the phantom image at 40 kVp tube voltage (a) 100% glandular tissue; (b) 70% glandular tissue; (c) 100% adipose tissue. The error bars indicate the range in CNR in each region of interest.

5.4.4 Summary and discussion

A breast tissue-equivalent mathematical phantom was simulated for digital mammography to resemble a physical tissue-equivalent phantom designed for mammography. The exposures estimated for different tube voltages and tube currents showed good agreement with the measured values (with a good correlation and slope of nearly one, as indicated in Figure 4.5 in Chapter 4). These estimated exposures were used for calibration of the incident energy spectrum for a given technique factor. The estimated analogue-to-digital conversion gain shows good agreement with values set manually. The additive noise due to dark signal and electronics was measured using digital images ac-

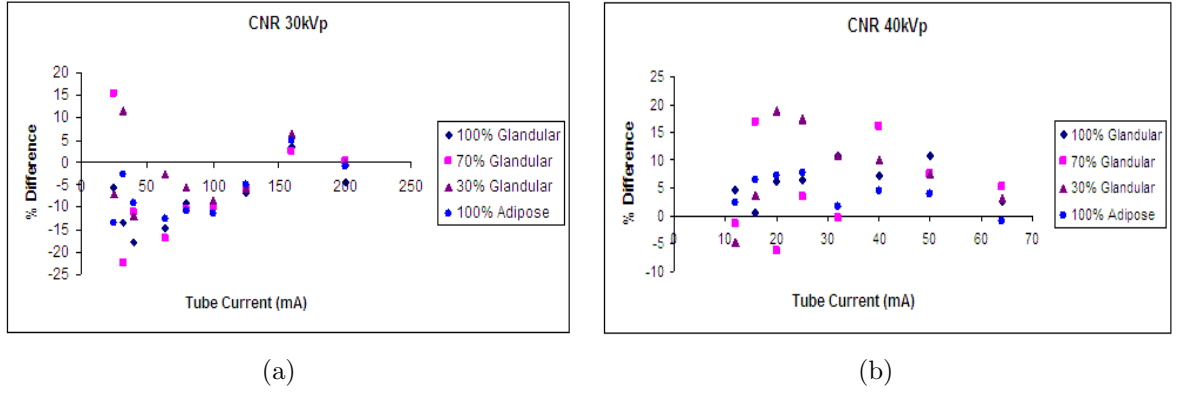


Figure 5.11: Differences between the measured and simulated CNR for different kVps as a percentage of their mean (a) 30 kVp; and (b) 40 kVp

quired with x-ray radiation off. Estimated CNR values depend on the estimation of pixel intensity and noise; hence higher differences are expected. Differences up to 22% are shown in Figure 5.11. The actual density and composition of the component regions of the physical phantom may play a role in producing larger differences, if the theoretical values used in the simulations misrepresent the physical values. The differences between the measured and simulated CNR are more pronounced in the 30% and 70% glandular tissue, which may indicate disagreement between the theoretical and physical composition of these regions. Our larger differences occur at lower tube current, similarly to those of (Doyle *et al.*, 2006), who reported differences in the measured and estimated CNR for chest radiography of up to approximately 20% as acceptable. As shown in Figure 5.6, Figure 5.9 and Figure 5.10, the contrast and contrast-to-noise show agreement between the physical and simulated images but the noise texture is different as shown in Figure 5.8; the simulated images show higher apparent granularity, which may affect lesion detectability in them.

This model estimates both qualitative and quantitative image quality parameters for detector design and optimization, and provides both quantitative and visual assessment of system performance. Simulation of structures in the mathematical phantom that exist in a physical phantom designed for mammography, allows direct comparison between simulated and acquired images for a given system and allows calibration of the mathematical phantom. The model also provides visual tools for the estimation of system resolution, which is limited by the image matrix size (maximum of 7 lp per mm for 12 bit image, 4069 x 4069 pixels). The model provides simulation tools which can easily be calibrated for different x-ray tubes and detectors. Most of the available simulated breast phantoms

(Bakic *et al.*, 2002; Bliznakova *et al.*, 2003; Hunt *et al.*, 2005; Dance *et al.*, 2005; Reiser *et al.*, 2006) are modelled to produce simulated mammograms that are visually similar to real mammograms, and need clinical breast images for model calibration and validation. Calibrating and validating the model for different mammography detectors then requires volunteer subjects to be exposed to radiation. These models also assume the only source of noise to be the natural fluctuation of the photon flux (quantum mottle), and neglect the additive noise due to the detector electronics which influences system performance. The following chapter presents model applications, namely evaluating different detector configurations and optimizing the x-ray spectrum.

5.5 Author's contribution

The author developed a computational tool titled *PhantomUser* (described in details in Appendix A) for simulating the physical tissue-equivalent phantom designed for mammography (Model 011A). This tool was developed for further system evaluation and optimization. The author performed the validation study described in this chapter.

Chapter 6

Applications of the Mathematical Breast Phantom

6.1 Introduction

Tissue equivalent breast phantoms have been used in several studies to investigate image quality in digital mammography and for the optimization of digital detectors (Smith *et al.*, 1998; Chakraborty and Eckert, 1995; Pachoud *et al.*, 2004; Khalifah *et al.*, 2006). These phantoms have been used to investigate image quality in a qualitative way by assessing the visualization of microcalcifications and tumour masses in the phantom image and in a quantitative way in terms of image quality parameters such as image contrast, spatial resolution, contrast-to-noise ratio (CNR) and detection efficiency.

This chapter presents the application of the mathematical phantom developed in the previous chapter to examine the effect of different detector configurations and for optimizing the x-ray spectrum by using a monochromatic x-ray beam. All programs were coded by the author using Matlab (Matlab, The MathWorks, Natick, MA).

6.2 Evaluating different detector configurations

Detector performance is influenced by several factors such as x-ray spectrum, spectrum filtration, type of scintillator, detector coupling efficiency and detector noise. The performance of a mammography detector can be measured by its ability to detect microcalcifications against a noisy background. A good mammography system is expected to detect a microcalcification as small as 100 to 200 microns (Smith *et al.*, 2004). The evaluation of the developed LSSM system using the physical tissue-equivalent phantom as described in Chapter 4, showed the ability of the system to detect microcalcifications of size $165\mu\text{m}$ at mean glandular dose (MGD) of 0.3mGy and the smallest microcalcifications ($130\mu\text{m}$) at MGD of 0.5mGy.

6.2.1 Method

Table 6.1 shows the technique factors used in examining the effects of different CCD coupling configurations and different levels of electronic noise on detector performance.

A CsI:TI scintillator (FOS, 1996) combined with a direct coupling detector, as used in the LSSM system, was compared to a CsI:TI scintillator-indirect coupling detector combination to evaluate the effect of direct coupling on system performance. Indirect coupling was achieved through a fibre optic taper.

The measured additive noise of the LSSM detector, with specifications as described in Table 6.1, was found to be 4850 electrons. In practice, the improvement of detector performance depends on the level of the electronic noise, which depends on the electronic design of the components of each x-ray imaging system, such as preamplifier design and signal processing electronics.

Table 6.1: Summary of LSSM system specifications, SDD is source-to-detector distance

	Parameter	Specifications
X-ray source		Varian Tube RAD70
	Focal spot	0.3mm
	Anode angle	16°
	Power rate	10kW
	Technique Setting	30kVp, 64 - 200mA
	Beam width at Detector	4mm
	Detector Exposure	70 - 200 μ Gy
SDD		650mm
Scanning speed		12 - 48 mm/s
Detector		Hamamatsu
Scintillator	CsI	150 μ m
Coupling efficiency	Direct coupling	0.08
	Indirect coupling	0.009
CCD		Hamamatsu
Additive noise		4850 e ⁻
Maximum data transfer rate		1ms
Pixel size		0.048mm

6.2.2 Results

Figure 6.1 shows the effect of using a direct coupling detector compared to coupling using a fibre optic taper. Figure 6.2 shows the improvement of system performance by reducing the additive noise, which will allow the system to detect the smallest microcalcifications (122 microns; bottom left in (a)). The bar plots in Figures 6.1 and 6.2 above each pair of images show the ratios of detector performance (image contrast, mean glandular dose (MGD), contrast-to-noise ratio (CNR), and system detection efficiency (DQE)). Values above unity represent an improvement in the image quality parameters or an increase in the mean glandular dose.

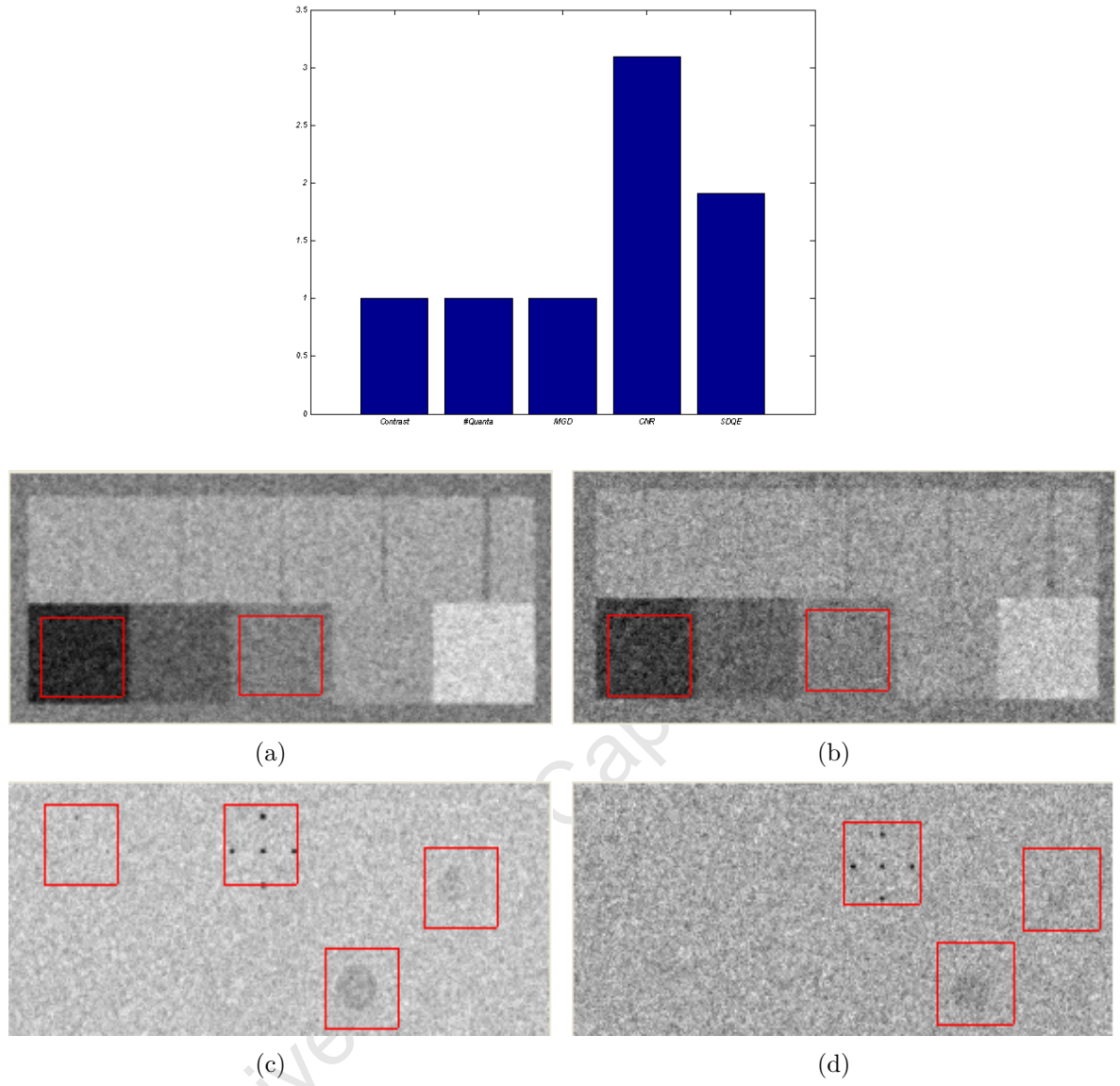


Figure 6.1: Image simulated using different optical coupling. Image of (a) step wedge using direct coupling; (b) step wedge using fibre optic taper; (c) microcalcifications and tumour masses using direct coupling; (d) microcalcifications and tumour masses using fibre optic taper. The bar plot indicates the improvement of output signal in terms of CNR and DQE at the same detector exposure, of direct coupling compared to indirect coupling.

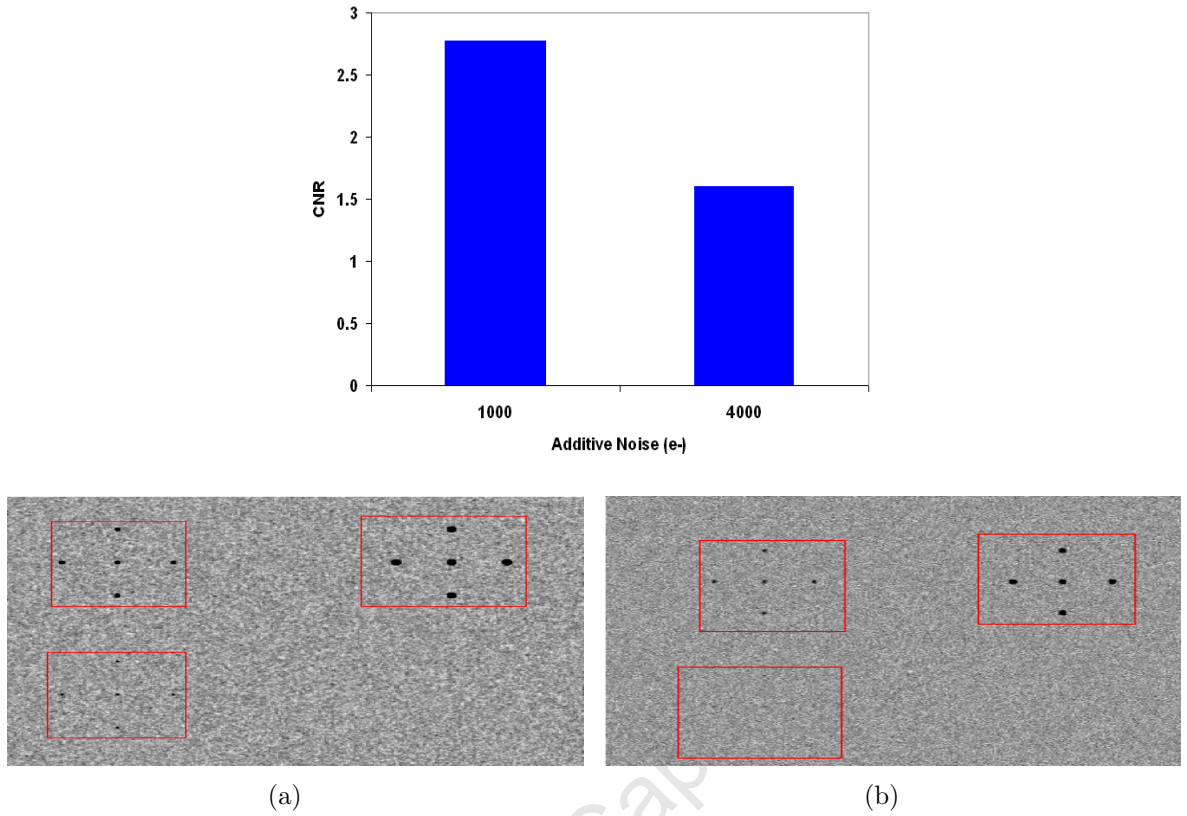


Figure 6.2: The effect of reducing the additive noise on detector performance. Simulated images using additive noise of (a) 1000 e- (b) 4000 e- . The bar plot indicates the improvement of detector performance in terms of CNR by reducing the additive noise (4000 e-).

6.3 X-ray beam optimization

6.3.1 Background

There are limitations to using a polychromatic x-ray spectrum, i.e. the presence of a wide range of x-ray energies, in mammography. Firstly, the low energy photons are mainly absorbed by the breast tissue and serve to increase the absorbed dose to the patient. Secondly, the higher energy photons increase the contribution of Compton scatter, which degrades the image quality. A narrow energy spectrum can provide better contrast and significant dose reduction (Boone and Siebert, 1994; Burattini *et al.*, 1995).

Using synchrotron radiation to produce a monochromatic beam (Burattini *et al.*, 1995; Johnston *et al.*, 1998; Moeckli *et al.*, 2000) is expensive. A new system producing a

quasi-monochromatic x-ray spectrum based on Bragg diffraction is available (Gambaccini *et al.*, 1995; Burattini *et al.*, 1995; Gambaccini *et al.*, 1997, 2001). This system consists of a tungsten anode x-ray tube and a graphite mosaic crystal array to filter the polychromatic beam and generate a monochromatic beam. Studies using the quasi-monochromatic beam showed similar image quality (contrast-to-noise ratio (CNR) and detective quantum efficiency (DQE)) with half the radiation dose to the patient as that delivered by conventional diagnostic x-ray imaging (Burattini *et al.*, 1995; Gambaccini *et al.*, 2001). The initial evaluation of a prototype digital mammography unit operating with a nearly monochromatic x-ray beam showed that the use of monochromatic slot-scan mammography resulted in correct identification of 93% of calcifications within a contrast-detail phantom (Diekmann *et al.*, 2004). This section describes the simulation of a monochromatic beam based on a filter produced by MXF technologies (MXF, 2006). The monochromatic source used to produce a narrow monochromatic beam from a polychromatic x-ray beam uses Bragg's law. The x-ray beam incident on a crystal will be reflected at a certain angle (θ); this is known as x-ray diffraction. For a given incident angle θ and a distance d between atomic layers in a crystal, the monochromatic tuned energy can be estimated from Bragg's equation:

$$E = \frac{12.4n}{2d\sin(\theta)} \quad (6.1)$$

where d is the distance between the lattice planes, n is an integer representing the diffraction order and θ is the angle between the x-ray source and the diffracting crystal or filter. The energy of the generated monochromatic beam depends mainly on the diffraction order n assuming the factor $2d\sin(\theta)$ is constant for parallel incident beams as shown in Figure 6.3. In reality, the photon beams from the x-ray source are not parallel and they have different incident angles; to account for the divergence of beams, the thickness of the crystal varies. The MXF filter tunes the x-ray spectrum to a particular energy suitable for the tissue being examined.

6.3.2 Method

The mathematical breast phantom was used to study the feasibility of using a monochromatic filter (MXF, 2006) for spectrum optimization in slot scanning mammography. The monochromatic beam was simulated using a Gaussian distribution according to the data provided by the MXF manufacturer (MXF, 2006). Two factors characterize the shape of

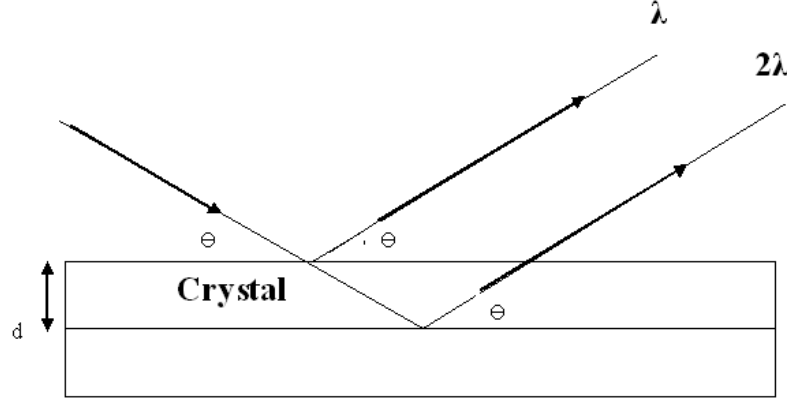


Figure 6.3: Bragg's law , x-ray diffraction

the monochromatic x-ray beam; the reflectivity of the filter and the spectrum resolution (FWHM). The reflectivity is a function of the roughness between the crystal layers and the way that they are fabricated. Using the data provided by the manufacturer (MXF, 2006), the reflectivity of the MXF filter as a function of energy can be estimated using the equation:

$$MXF_{Ref} = 80 - 0.5MU \quad (6.2)$$

MU is the tuned energy for the monochromatic beam.

The FWHM, which represents the beam resolution (beam width), depends on the roughness between the layers. The measured FWHM for the MXF monochromatic beam is 10% for the energy range between 20 and 60kVp as provided by the manufacturer. The calculated beam width in keV related to this value is given by:

$$Width_{keV} = \frac{FWHM \times MU}{100} \quad (6.3)$$

The width of a Gaussian distribution at half maximum is equal to 2.355σ . The standard deviation in terms of the beam width can then be estimated using the relation:

$$\sigma = \frac{Width_{keV}}{2.3456} \quad (6.4)$$

Figure 6.4 shows the calculated monochromatic beams obtained from the polychromatic beam for different tuned energy levels using data provided by MXF and the above equations. The detective quantum efficiency (DQE) was estimated using cascaded linear

analysis (Siewerdsen and Jaffray, 2000). Mean glandular dose (MGD) was calculated using Boone's fit equations (Boone, 2002).

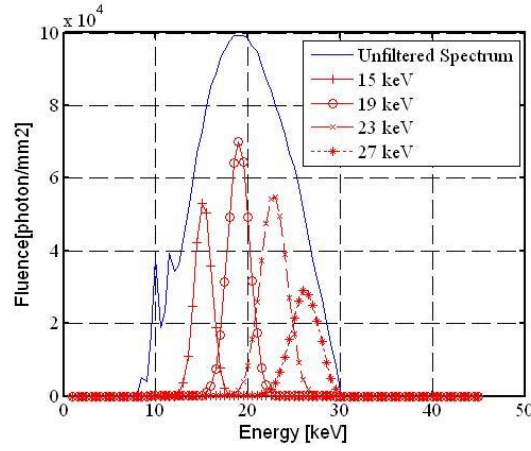


Figure 6.4: The estimated MXF monochromatic beams at different tuned energy levels and the unfiltered spectrum.

6.3.3 Results

Figures 6.5(a) and 6.5(b) show the effect of a monochromatic beam compared to that of a polychromatic beam on image contrast and detective quantum efficiency (DQE).

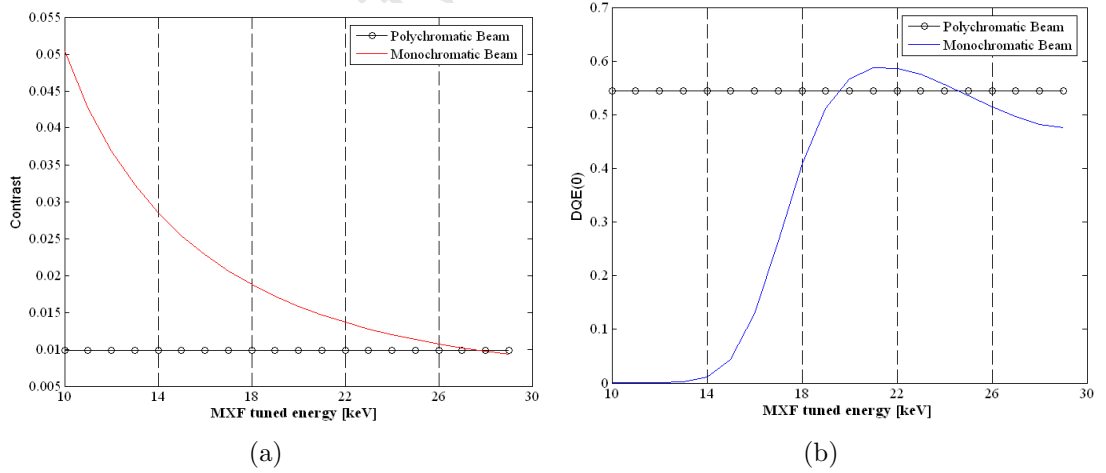
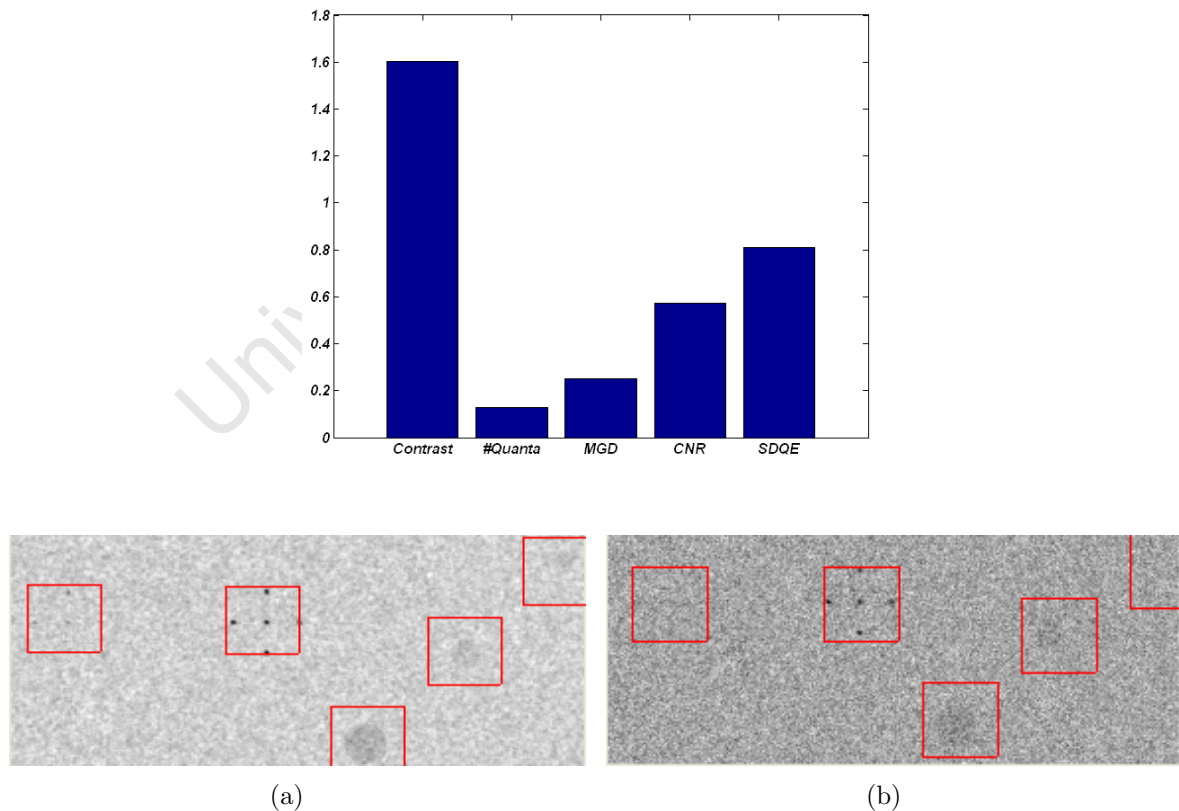
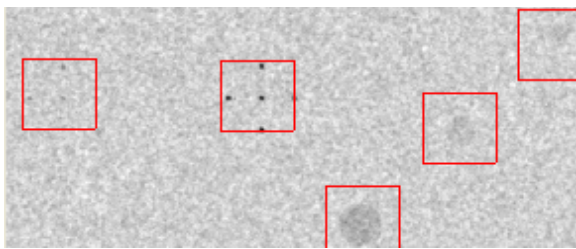
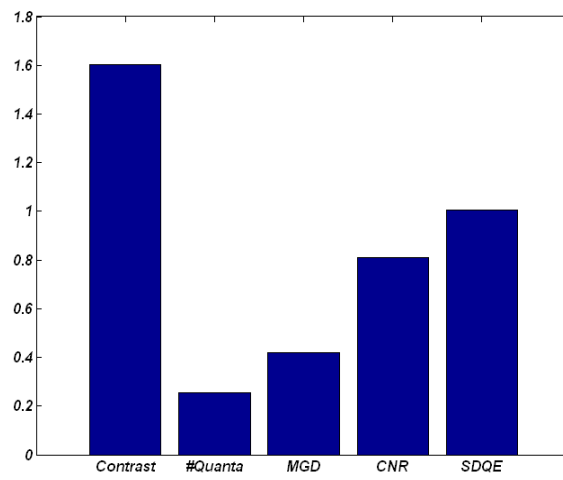


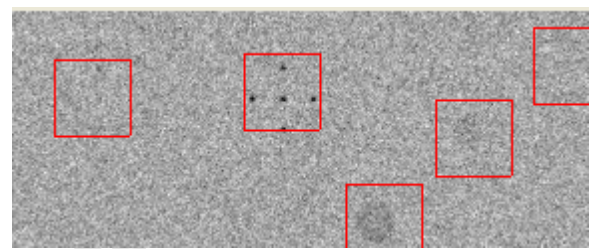
Figure 6.5: The effect of monochromatic beams at different tuned energy levels on the image contrast and the system DQE. The image quality parameters were calculated using the simulation model (*MammoModel* described in Appendix A).

Figure 6.6 shows the effect of using the MXF monochromatic beam on detector performance in terms of image contrast and dose reduction. The bar plots in Figure 6.6 above each pair of images show the ratios of monochromatic to polychromatic performance (image contrast, mean glandular dose (MGD), contrast-to-noise ratio (CNR), and system detection efficiency (DQE)); values above unity represent an improvement in the image quality parameters or an increase in the mean glandular dose. A monochromatic x-ray beam tuned to energy in the range between 19 and 21keV (a range shown in Figure 6.5 to produce superior contrast and DQE to those of a polychromatic beam) will result in a 60% improvement in contrast as shown in the first three bar plots. Using the narrow spectrum will also result in a dose reduction and a consequent reduction in photon flux. In order to achieve the same contrast-to-noise ratio as for a polychromatic beam, an increase in scanning time or in tube loading is required for the monochromatic beam. For the monochromatic beam (10% FWHM), tripling the scanning time will result in the same contrast-to-noise ratio with a 50% dose reduction as shown in Figure 6.6(f). Using a wider monochromatic beam (e.g. 25% FWHM) can produce the same image quality with less tube loading as shown in Figure 6.6(h).

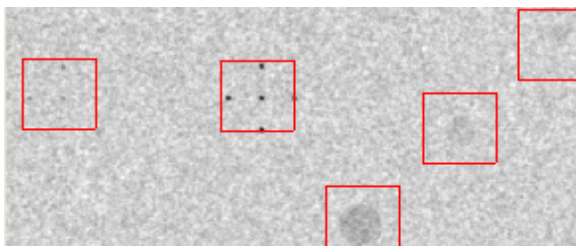
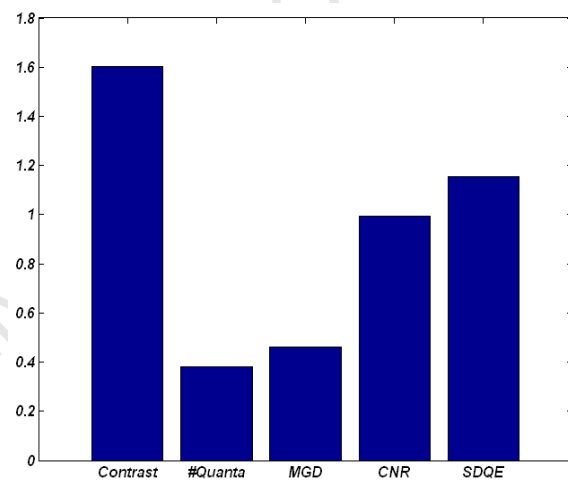




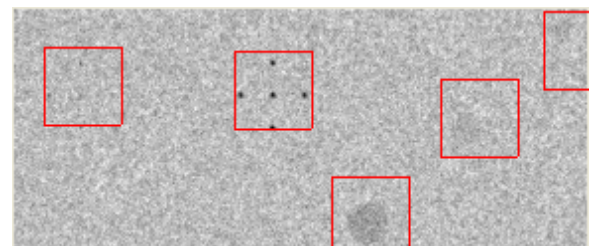
(c)



(d)



(e)



(f)

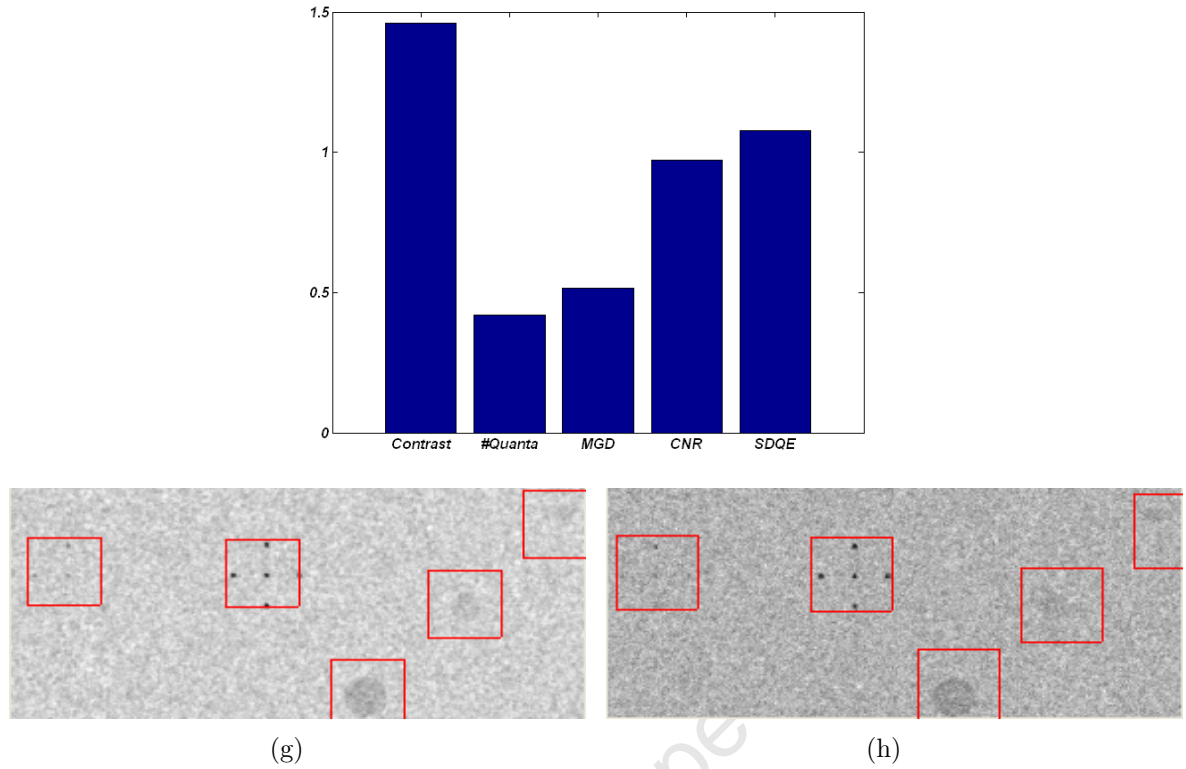


Figure 6.6: The image performance of the polychromatic (left) and the monochromatic (right) beams at tube voltage 50kVp and tube current 100mA - the polychromatic beam is the unfiltered spectrum shown in Figure 6.4 and the monochromatic, is a beam tuned to 19 keV; (a) and (b) the performance using the same technique factors; (c) and (d) doubling the scanning time of the monochromatic beam; (e) and (f) tripling the scanning time of the monochromatic beam; (g) and (h) doubling the scanning time and using FWHM of 25% rather than 10% for the monochromatic beam.

6.4 Summary and discussion

The mathematical phantom was used to examine the effect of different detector configurations and different levels of electronic noise on detector performance. The use of a CsI scintillator directly coupled to the charge coupled device (CCD) increases the efficiency of the detector and hence improves the detector performance compared to using indirect coupling. Improving the detector performance in terms of electronic noise reduction and coupling efficiency will lead to a significant dose reduction.

The optimization of the x-ray spectrum using the MXF monochromatic filter instead of a polychromatic filter results in better contrast and the same image quality (CNR and DQE) with a significant dose reduction. The performance of the monochromatic filter depends on the tuned energy and the beam width (FWHM). The optimum tuned energy is a trade-off between the image contrast and photon flux reaching the detector

(detector exposure), which affects the detection efficiency of the detector. Figures 6.5(a) and 6.5(b) show the calculated image contrast which decreases with energy while the detection efficiency increases. The optimum tuned energy for imaging an average breast (4cm thickness), i.e. that which gives the highest detection efficiency, was found to be in the range between 19keV and 21keV. A significant mean glandular dose reduction was found when using the monochromatic beam rather than the polychromatic beams. This reduction is combined with a three-fold increase in tube load to achieve the same signal-to-noise ratio when using 10% FWHM. This increase in the tube load can be reduced by using a wider monochromatic beam (e.g. 25% FWHM).

The mathematical phantom applications discussed in this chapter show the benefit of using this tool in system optimization and improvement of image quality. The resulting dose reduction may reduce patient discomfort that results from painful breast compression.

6.5 Author's contribution

The author performed the studies and interpreted the results described in this chapter.

Chapter 7

Photon Counting Threshold Optimization in Mammography

7.1 Introduction

Photon counting detectors are a promising approach in digital mammography (Cahn *et al.*, 1999). In energy integrating detectors the signals from all interacting photons in a pixel are integrated to form the pixel value, while in photon counting detectors all the detected photons are assigned a weighting factor of 1. Another promising approach is the integration of an analogue-to-digital converter into the detector for each image pixel. This would allow detection of each single photon and also measuring its energy. In energy weighting detectors, each photon is weighted by a factor proportional to its energy; the higher the weighting the less the contrast information. The total signal is equal to the sum of the weighted signal in each energy channel (Giersch *et al.*, 2004).

$$S_{total} = \sum S_i \omega_i \quad (7.1)$$

The weighting factor ω_i is proportional to the signal of an object embedded in a uniform material as seen in Figure 7.1.

$$\omega_i \propto [1 - e^{-(\mu_1(E_i) - \mu_2(E_i))x}] \quad (7.2)$$

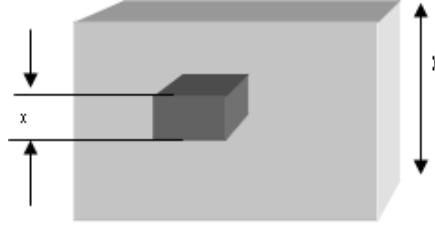


Figure 7.1: Small object embedded on a uniform material

For a low-contrast object, the weighting factor is proportional to the difference in attenuation coefficient between the object and the uniform material.

$$\omega_i \propto \mu_1(E_i) - \mu_2(E_i) \quad (7.3)$$

For mammography, the broad-spectrum detective quantum efficiency (DQE_{task}) theory has recently been extended to take energy weighing into consideration (Marchal, 2005). The DQE is given by:

$$DQE_{task} = \frac{SNR^2}{SNR_{max}^2} \quad (7.4)$$

Where SNR is the broad-spectrum signal-to-noise ratio, for a specific task; SNR_{max} is max value of the broad-spectrum SNR of an ideal detector that preserves all the information without adding any noise.

For a low-contrast object detected under broad-spectrum imaging conditions, the broad-spectrum DQE is given by (Marchal, 2005):

$$DQE^{bs}(f) = \frac{[\int_{\varepsilon} G(E) \Delta\mu(E) q_b(E) T(f, E)]^2 dE}{NPS(f) \int_{\varepsilon} \Delta\mu^2(E) q_b(E) dE} \quad (7.5)$$

where f is the spatial frequency and $q_b(E)$ is the x-ray fluence incident on the detector after breast attenuation, over the energy spectrum ε . $\Delta\mu$ is the difference of linear attenuation between the background and the microcalcifications to be detected, and is proportional to the photon weighting factor (ω_i). The energy-dependent transfer function $T(f, E)$ and the noise power spectrum $NPS(f)$ characterize the light spreading and quantum noise. The large-area detector gain $G(E)$ of the photon counting detector is represented by a step function with a rising edge determining the photon counting threshold. $\int_{\varepsilon} \Delta\mu^2(E) q_b(E)$ is the maximum signal to noise ratio (SNR_{in}) for an ideal detector.

The influence of scatter reduction methods such as slot scanning geometry and additive noise are usually treated independently of the detector performance. Introducing the contributions of the primary radiation, scattered radiation and additive electronic to the NPS leads to the effective broad-spectrum DQE (Marchal, 2005).

$$DQE_{eff}^{bs}(f) = \frac{t^2 [\int_{\epsilon} G(E) \Delta\mu(E) q_b(E) T(f, E)]^2 dE}{NPS_P(f) + NPS_S(f) + NPS_{add}(f)} \times \frac{1}{\int_{\epsilon} \Delta\mu^2(E) q_b(E) dE} \quad (7.6)$$

7.2 Single photon counting and detection thresholds

Photon counting detectors have a number of advantages over charge integrated detectors. These advantages are: the improvement of image quality, the dose efficient rejection of scattered radiation, and effective rejection of electronic noise and Swank noise. The detection and counting of each photon in the radiation beam, would allow energy resolved x-ray imaging. The photon counting detector assigns a more optimal energy-weighting factor (equal to 1) to the detected photons, as opposed to the charge integrating detector, where the weighting factor is approximately proportional to the energy of the photon (Tapiovaara and Wagner, 1985; Cahn *et al.*, 1999; Giersch *et al.*, 2004; Marchal, 2005).

The selection of an optimal threshold is of interest when designing an imaging system based on a photon counting detector. Although the selection of an optimum threshold can reduce photon double counts due to charge-sharing effects (the charge from a photon interaction at a certain position is divided across adjacent channels) it also reduces the SNR due to the rejection of useful primary incident x-rays. We recently proposed a novel approach for the selection of detection thresholds in single-photon counting sensors developed for digital mammography (Marchal *et al.*, 2006). Based on the approach proposed by Tapiovaara and Wagner (1985) and applied to mammography by Cahn *et al.* (1999), the zero-frequency DQE of an ideal photon counting detector for a low-contrast object in broad-spectrum imaging conditions is as follows:

$$DQE(0) = \frac{[\int_{\epsilon} G(E) \Delta\mu(E) q_b(E) T(0, E)]^2 dE}{NPS(0) \int_{\epsilon} \Delta\mu^2(E) q_b(E) dE} \quad (7.7)$$

The detector efficiency and gain are equal to 1 for ideal photon counting detectors. The figure of merit (FOM) can estimate the system performance in terms of a benefit to risk

ratio, which is related to the ratio of SNR_{max} to the mean glandular dose (Dg).

$$FOM = \frac{1}{Dg} \times DQE(0) \times \int_{\epsilon} \Delta\mu^2(E) q_b(E) dE \quad (7.8)$$

This chapter presents a computer model developed to assist in implementation of novel algorithms developed by Marchal *et al.* (2006) for the selection of detection thresholds in single-photon counting in digital mammography. The image quality parameters detective quantum efficiency (DQE) and figure of merit (FOM) were used to estimate the effect of the selection of optimum detection thresholds in the performance of digital mammography using different imaging tasks and different x-ray spectra. All algorithms were coded by the author and Julien Marchal using Matlab (Matlab, The MathWorks, Natick, MA). The code was implemented by the author in a MATLAB graphical user interface (GUI) titled *OptimumThreshold* (see Appendix A).

The author of this thesis is a co-author of the paper by Marchal *et al.* (2006), and this chapter reflects his contribution to the published paper.

7.3 Methods

The image quality parameters detective quantum efficiency (DQE) and figure of merit (FOM) were calculated for various threshold energies using the developed novel theoretical approach described above. The effect of different technique factors, target-filter combinations, and external filtration can be examined. The x-ray spectra, imaging task, mean glandular dose (MGD) and scatter to primary ratio (SPR) were simulated using published data from the literature and the algorithms discussed in Chapter 3.

7.3.1 X-ray spectra

Two target-filter combinations (molybdenum-molybdenum and tungsten-aluminium) were investigated. X-ray spectra were generated using published data (Boone, 1997). A source-to-image distance of 650mm and an air gap of 12mm were considered with an infinitely small x-ray focal point in order to ignore the geometric blurring effects. The x-ray spectra were modified for an internal filtration of 0.76mm beryllium window

and a set of external filters namely molybdenum, rhodium and aluminium. The filter attenuation coefficients were computed using the program Xcom (Berger *et al.*, 2007).

7.3.2 Detector parameters

The quantum efficiency of the detector was assumed to be unity for an ideal photon counting detector. The only source of detector noise is quantum noise, governed by Poisson statistics. The gain $G(E)$ of the photon counting detector is represented by a step function with a rising edge determining the photon counting threshold.

7.3.3 Imaging Task

The imaging task consists of identifying a 100 μm microcalcification (calcium hydroxyapatite) of density $3.3\text{g}/\text{cm}^3$ in breast tissue of a radius 8.5 cm and different thicknesses between 2 and 8cm. The breast tissue is modelled as 50:50 glandular/adipose tissue of uniform density and composition. The attenuation coefficients of breast tissue and microcalcifications were computed using the program Xcom (Berger *et al.*, 2007).

7.3.4 Mean glandular dose and scatter-to-primary ratio

The mean glandular dose for different breast compositions and thicknesses was calculated from the incident photon fluence using the fit-equation for mono-energetic x-ray beams (Boone, 2002). Scatter-to-primary ratio (SPR) for Mo:Mo and W:Al spectra was calculated using fit-equations obtained from Monte-Carlo simulations (Boone *et al.*, 2000). The SPR for full field digital mammography is used in the model of the photon counting detector (Equation 3.9). The calculated SPR shows little dependance on the air gap between 0 and 30mm and on the x-ray spectrum.

7.4 Results

Using the *OptimumThreshold* model, the influence of the detection threshold on the DQE and FOM for the two combinations W:Al and Mo:Mo spectra for 50:50 glandular/adipose tissue and breast of thickness 2, 4, 6 and 8cm was examined. Figures 7.2 and 7.3 show the effect of the detection threshold on DQE and Figures 7.4 and 7.5 show the effect of detection thresholds on the FOM obtained with ideal photon counting detectors for W:Al(0.5mm) and Mo:Mo(0.03mm) spectra at various kVp settings.

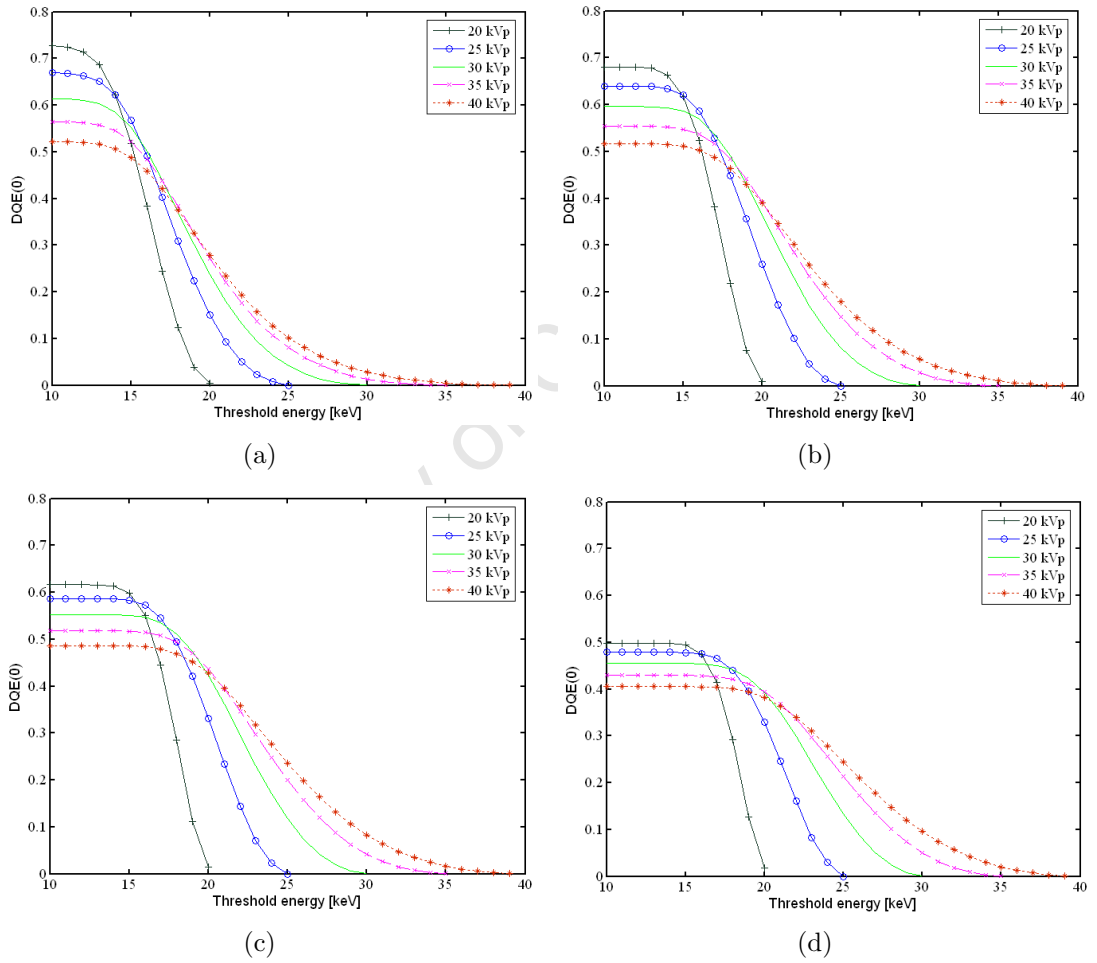


Figure 7.2: Effect of detection energy threshold on the DQE at zero spatial frequency for W:Al(0.5mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.

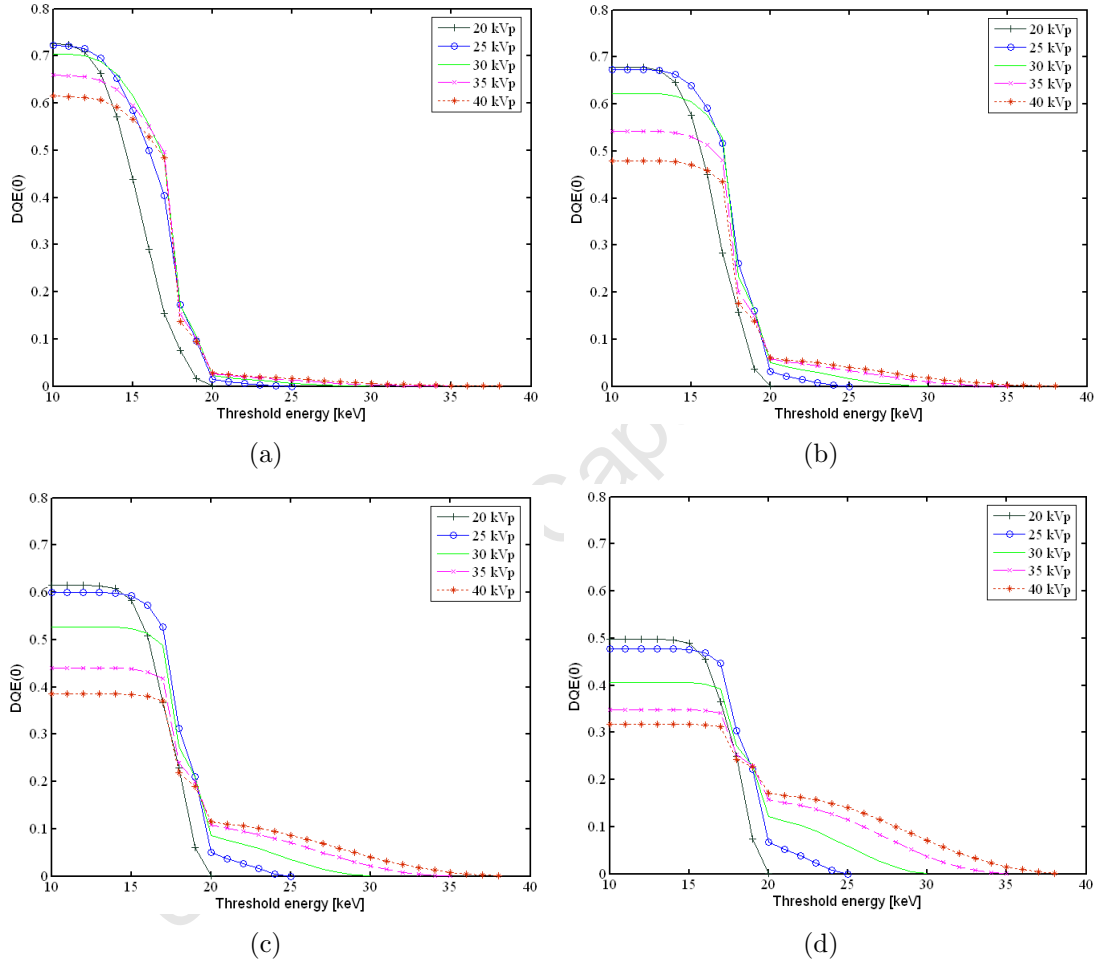


Figure 7.3: Effect of detection energy threshold on the DQE at zero spatial frequency for Mo:Mo(0.03mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.

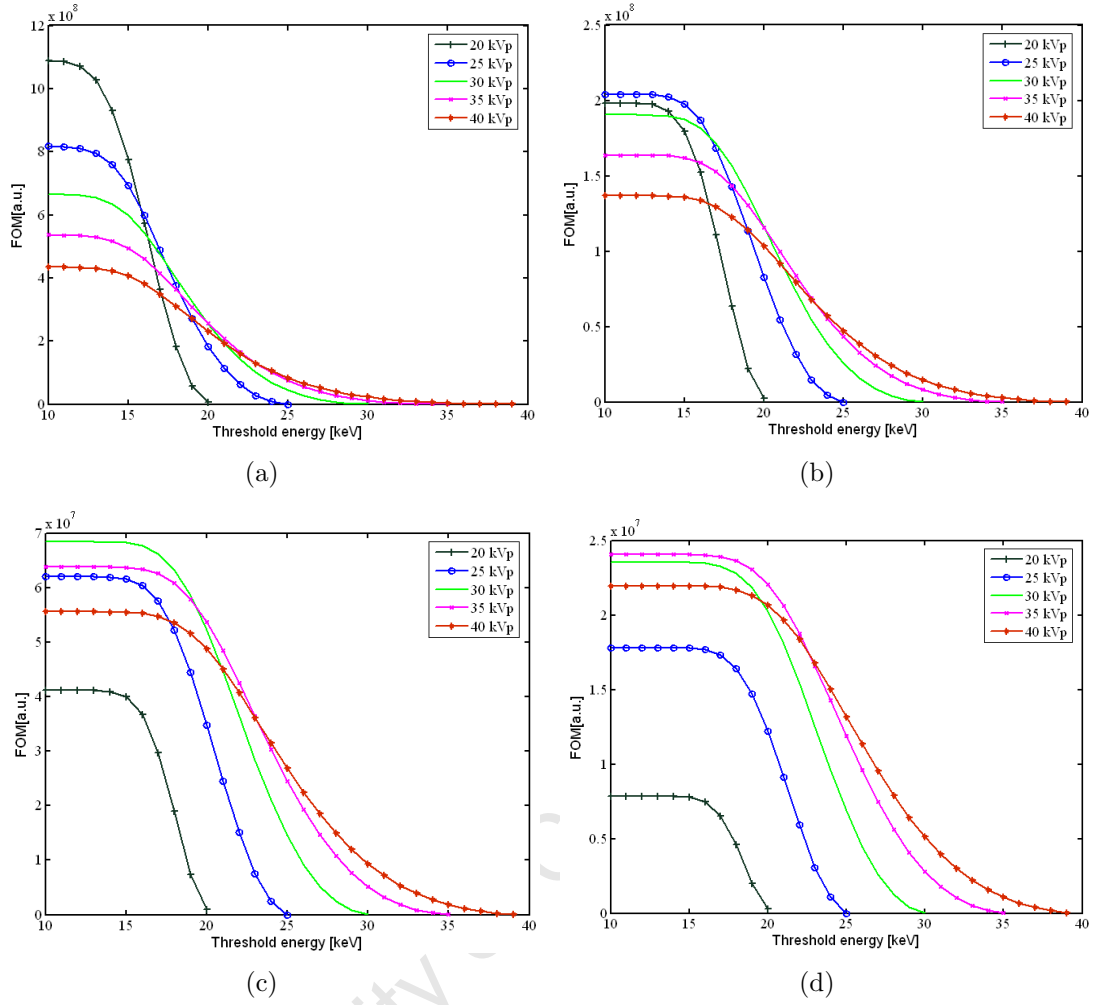


Figure 7.4: Effect of detection energy threshold on the FOM for W:Al(0.5mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.

The effect on the FOM of the external filter material and different detection thresholds for tungsten spectra, for 50:50 glandular/adipose tissue, and breast thickness of 8 cm at various kVp settings, is shown in Figure 7.6.

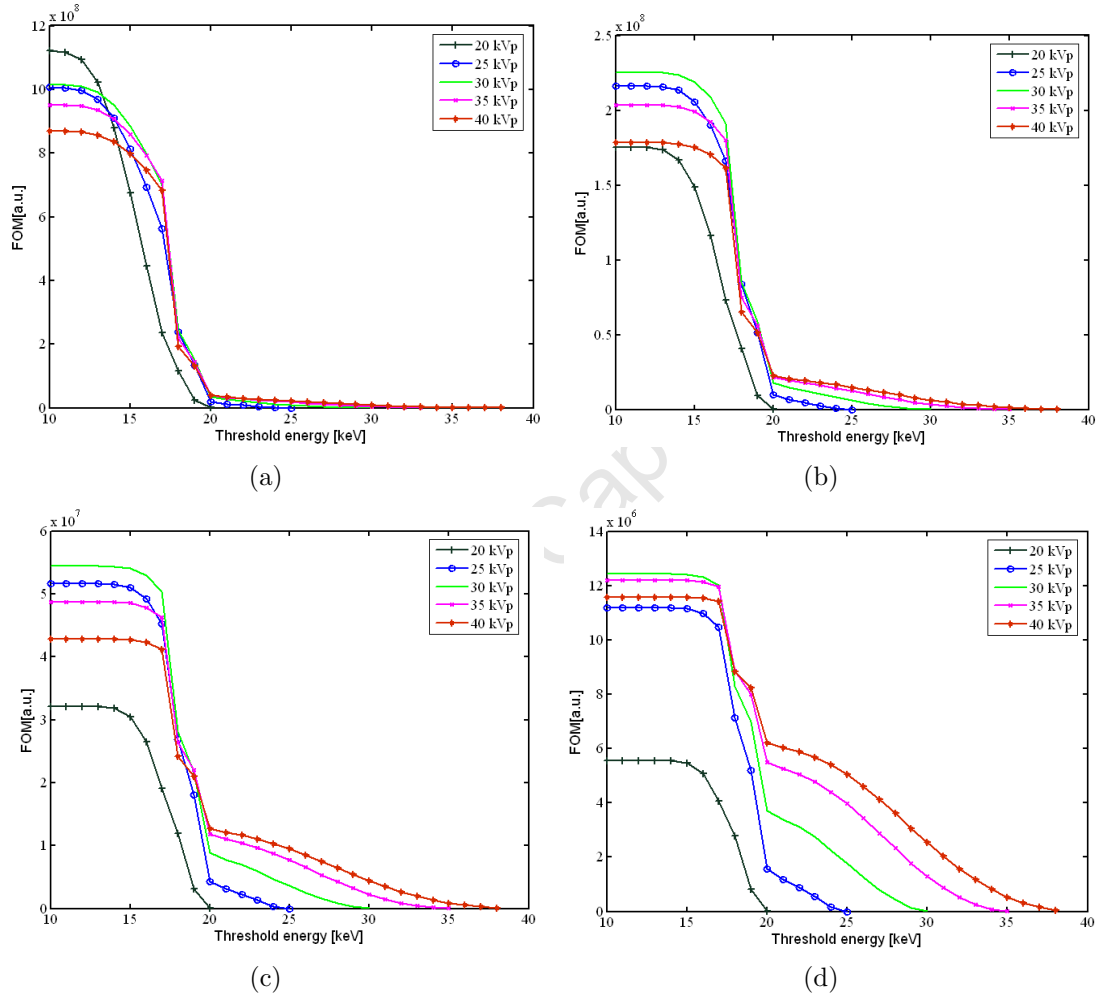


Figure 7.5: Effect of detection energy threshold on the FOM for Mo:Mo(0.03mm) spectra at various x-ray tube peak voltages for a 50% glandular compressed breast of thickness (a) 2 cm, (b) 4 cm, (c) 6 cm and (d) 8 cm.

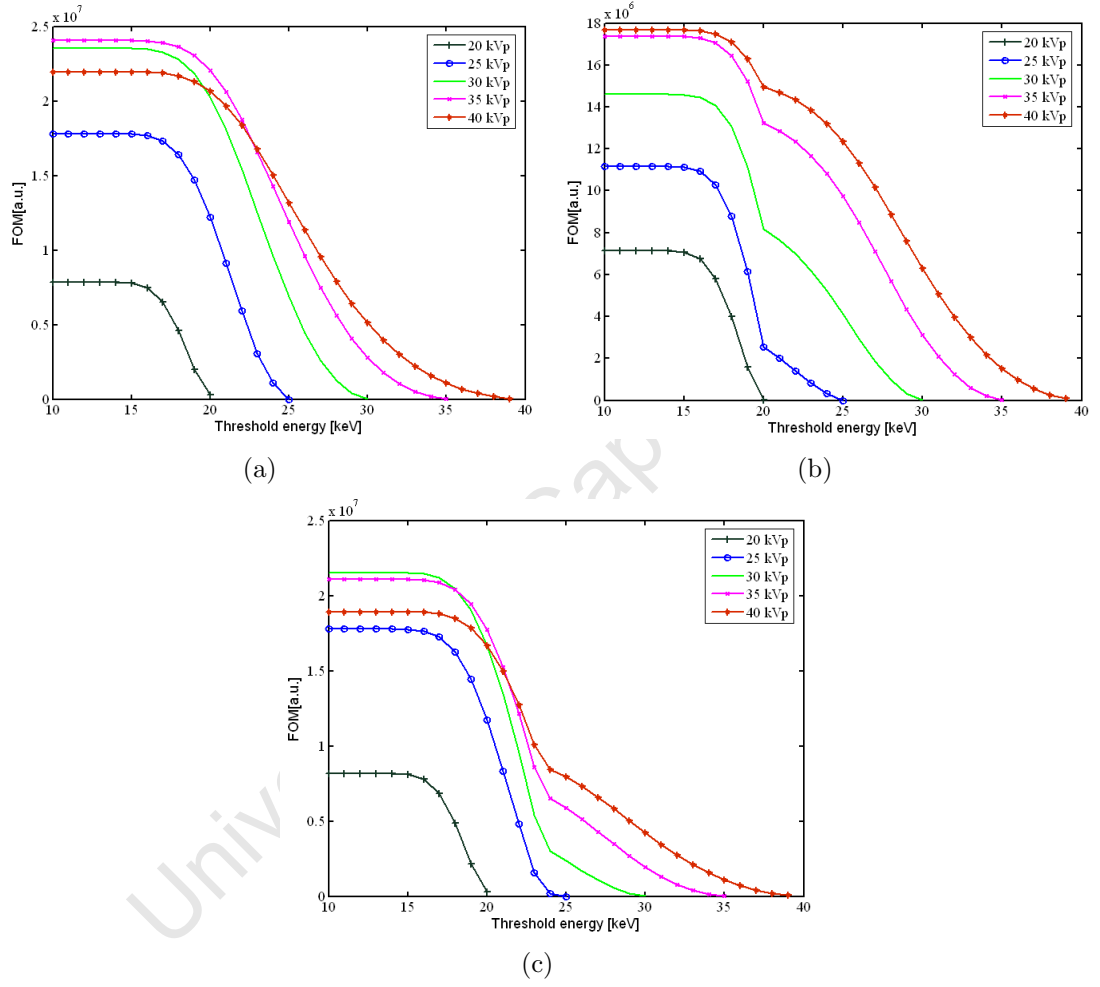


Figure 7.6: Effect of detection energy threshold on the FOM for tungsten spectra at various x-ray tube peak voltages for a 50% glandular 8cm compressed breast, the spectrum filtered with (a) Al (0.5mm), (b) Mo (0.03mm), (c) Rh (0.03mm).

7.5 Summary and discussion

The computational tool *OptimumThreshold* presented in this chapter was developed to assist in the implementation of the novel algorithms developed by Marchal *et al.* (2006), and implemented for the purpose of digital mammography optimization. Using a monochromatic beam with an optimum tuned energy or selecting an optimum detection threshold for a polychromatic beam can play an important role in spectrum optimization and reducing charge-sharing effects. With a monochromatic beam, increasing or decreasing the optimum tuned energy will affect the DQE and hence the performance of digital mammography. Figures 7.2 and 7.3 indicate that the system performance in terms of DQE degrades by increasing breast thickness or detection threshold. This is due to the increase in the amount of scattered radiation, which increases with breast thickness. With a threshold smaller than 20keV for thinner breast thickness, where low tube voltage is used, the Mo:Mo target-filter combinations will cut the useful information in the higher part of DQE curve, while in larger breast thickness, where high kVps are selected, there is still useful information in the higher part of the DQE curve. This indicates that higher detection thresholds can be implemented for a thick breast without compromising imaging performance (Marchal *et al.*, 2006). Using the W:Al spectrum rather than Mo:Mo provides a smoother DQE curve especially for larger breast thickness, due to the absence of sharp peaks in the W:Al spectrum, as indicated in Figure 7.2 (Marchal *et al.*, 2006). Tungsten tubes offer greater output power and better heat dissipation properties, making them suitable for slot scanning mammography especially when dealing with thicker breasts.

Introducing the absorbed dose in the breast into the DQE leads to another more meaningful image quality parameter known as figure of merit (FOM), which accounts for the dose efficiency as well as the risk factor due to the radiation dose. Figures 7.4 and 7.5 show the effect of detection thresholds on the FOM obtained with ideal photon counting detectors for W:Al and Mo:Mo at various kVp settings for different breast thicknesses. These results were found to be consistent with the technique normally used in mammography, namely increasing kVp to produce higher-quality images of thick compressed breasts.

Figure 7.6 shows the effect of detection thresholds on the FOM obtained for a tungsten x-ray spectrum filtered with various attenuating materials used commonly in mammography (aluminium, molybdenum, and rhodium). The 0.5mm Al filter shows smoother

variations due to the absence of sharp peaks compared to the other filters. The k-edge of molybdenum (20.0 keV) and rhodium (23.2 keV) acts as an upper edge, where the DQE decreases rapidly when the threshold passes over these energies. The effect of detection thresholds on the performance of the mammography detector was studied using DQE and FOM at zero spatial frequency; for realistic detectors, the analysis should include all spatial frequencies.

7.6 Author's contribution

The author developed the computational tool *OptimumThreshold* discussed in this chapter, with assistance from Julien Marchal, who developed some of the theory in Marchal (2005). The author's contribution to the paper Marchal *et al.* (2006), of which he is a co-author, and which is cited in this chapter, is the computational tool and the results of applying it. Additional analysis, beyond what appears in Marchal *et al.* (2006), was done by the author for this chapter.

Chapter 8

Conclusions and Recommendations

This work formed part of an NIH-funded project entitled “Low dose digital X-rays for mammographic screening” conducted in the MRC/UCT Medical Imaging Research Unit to develop a prototype linear slot scanning mammography system (LSSM). Computational tools were developed for the design and optimization of the LSSM system.

A mathematical model entitled *MammoModel* was developed for obtaining the design specifications and system requirements of the LSSM system. The components of the LSSM were selected based on the simulated image quality and the required system specifications for mammography. The LSSM system was built having a similar linear scanning configuration as the Lodox Statscan full body slot scanning radiography system, but adapted for mammography using a mammography x-ray tube and a high performance detector mounted on a C-arm. The system performance was evaluated using a physical breast tissue-equivalent phantom designed for mammography.

System evaluation showed that the LSSM system approaches the resolution of analogue mammography with much better contrast and object detectability; equivalent image quality in terms of image contrast and resolution at a significant dose reduction compared to computed radiography (CR); and good contrast and much better resolution at equivalent dose compared to a digital mammography system.

As indicated in Chapter 4, the measured system resolution using line pair tools and an edge detection method showed a good resolution of nearly 9lp/mm. This gives the LSSM system superiority over some existing mammography systems and also confirms

the fact that the linear slot scanning system can offer a good resolution as well as high efficiency and low dose to the patient. Comparing the LSSM system, which uses a linear slot scanning configuration, with the Fischer system (Senoscan) (Tesic *et al.*, 1999), which uses a fixed source slot scanning configuration, the LSSR showed a significant dose reduction.

The measured additive noise was found to be 4850 electrons, which is considered high compared to that of existing mammography systems. The smallest microcalcifications of $130\mu\text{m}$ (the expected microcalcifications to be detected in a state of the art mammography system) were detected at this level of noise at a mean glandular dose of 0.5mGy.

The LSSM system shows a detective quantum efficiency (DQE) of 0.56, which is good compared to that of current digital mammography systems using the indirect detection method, but it is less than that of systems using direct and photon counting detectors.

Although slot scanning has a limitation regarding the photon flux, the improvement in image contrast and the reduction in dose have an impact on image quality, radiation hazard, and patient comfort. This makes linear slot scanning a promising technique for digital mammography.

A mathematical model entitled *PhantomUser* was developed for examining different detector configurations and for spectrum optimization using a monochromatic filter. Validation of the mathematical phantom showed a good agreement (within 95%) of values generated using the phantom with measured values, in estimation of the pixel intensity, the image contrast, and detector noise. This allows the phantom to be used in optimization of the LSSM system for better image quality a spectrum optimization showed the feasibility of using a monochromatic filter for image contrast improvement but with a reduction in photon flux and hence a reduction in the signal-to-noise ratio. This can be overcome by using a wider filter bandwidth.

A mathematical model entitled *OptimumThreshold* was developed to study the effect of energy thresholds and the anode-filter combination on photon counting detector performance. The results and analysis showed that the detector performance in terms of DQE at zero spatial frequency improved by decreasing the energy threshold. The W:Al anode-filter combination was found to be suitable for slot scanning mammography offering greater output power and better heat dissipation, especially when dealing with thicker breasts.

Future work: Despite the fact that evaluation of the prototype LSSM system using the breast phantom is not sufficient for the system to be used in clinical studies, the system has promising performance in terms of spatial resolution and detection efficiency over the existing mammography systems to which it was compared. Using custom-made detector electronics optimized for mammography, will lead to an improvement in the performance of the LSSM system. The improvement of detector electronics can be achieved by using high speed electronics for high speed data acquisition and collection while minimizing the additive electronic noise by cooling the detector or by using less noisy components. High speed data acquisition will result in shorter scanning time to reduce motion blurring artifacts. Although the LSSM system still needs validation on an appreciable patient base, it holds great promise as a tool for low dose and high resolution mammography.

Appendix A

Slot Scanning Mammography: MATLAB Graphical User Interface

A.1 Introduction

MATLAB (Matlab, The MathWorks, Natick, MA) graphical user interfaces (GUIs) were created for estimation of the image quality parameters, evaluation of detector configurations, and modelling and validation of a mathematical tissue-equivalent breast phantom for digital mammography. The majority of algorithms described in this thesis are implemented as functions in the GUIs, which were coded by the author. Julien Marchal contributed to the *OptimumThreshold* tool, as indicated in Section 7.6.

This documentation describes the use of GUIs but assumes that the user is familiar with MATLAB. Included on the CD that accompanies this thesis are three main script files (M-Files) necessary to run three graphical user interfaces: MammoModel.m, PhantomUser.m, and optimumthreshold.m. The user can run these programs through a main graphical user interface using the M-file MammoMain.m, but the other files must be in the same working directory or on the same MATLAB path.

A.2 Software Requirements

These GUI can run only in MATLAB. No extra toolboxes are needed. A high speed processor computer (not less than 2GHz) is recommended.

A.3 Basic Usage

The CD provided with this thesis contains the Mammography folder and the readme file.

- (1) Copy the Mammography folder and its contents to a local hard drive.
- (2) Launch MATLAB.
- (3) Set the MATLAB working directory to the Mammography folder.
- (4) Open the file MammoMain.m in the Editor.
- (5) Run MammoMain.m, the user will be allowed to select one of the three buttons shown in Figure A.1 below:

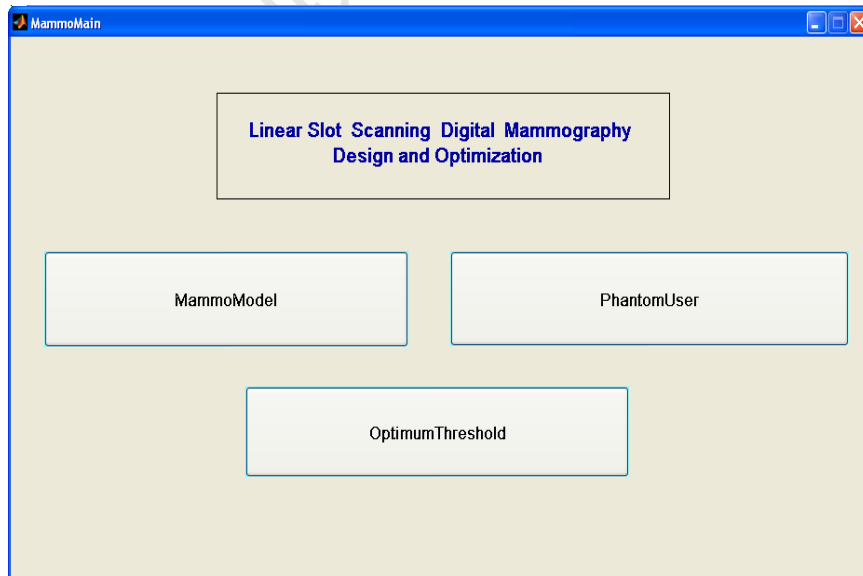


Figure A.1: The main graphical user interface (GUI) program.

MammoModel button. This allows the user to open the MammoModel GUI.

PhantomUser button. This allows the user to open the mathematical phantom GUI.

OptimumThreshold button. This allows the user to open the OptimumThreshold GUI.

A.4 MammoModel GUI

A GUI titled *MammoModel* was created for modelling the performance of a slot scanning mammography system. This GUI was also used for the derivation of system requirements for slot scanning mammography. Figure A.2 shows the GUI for design and optimization of slot scanning digital mammography. The GUI includes several functions for:

- Generation of x-ray energy spectrum.
- Applying an image task.
- Input of the detector parameters.
- Detector output parameters.

MammoModel
Untitled 4

X-Ray Spectrum
Fluence(photon/mm² x 10⁶) vs keV
Plot Spectrum
Clear Plot
Hold on
Select Color
g

X-RAY TUBE
Tube
Focal Spot
Focal Spot Size
Inherent Filtration
Anode Material

Dose(mGy)
8.76
Spectrum Calibration
Spectrum Calibration

TECHNIQUE FACTORS
Tube voltage & current
25 kVp
50 mA
Scan Speed
Quarter
37.5 mm/s
Added Filtration
0.5 Al(Z=13) mm
Collimator width
0.8 mm
Beam width
5.2 mm
Air gap
40 mm
Source-Detec Dist
650 mm
Source-Collim Dist
100 mm
Generate Spectrum
Apply filter

DETECTOR PARAMETERS
Scintillator Material
CsI
150 um
Time Decay
1 us
energy/lightphoton
18 eV
Swank Factor
0.98
Quantum Efficiency
0.80847
Conversion gain
1390 quanta/x-rays
CCD Quantum Efficiency
0.35
FOT Parameters
Trans_Eff_FOT
0.14969
Demagnification
Pixel Size(mm)
0.05
Fill Factor
0.74
FOT transmission
0.6
Pixel Bin
1
FOPtoFOT Coupling
0.45
Additive Noise Parameters
Coupling Efficiency
0.01
Additive Noise
4192

IMAGING TASK
Breast Thickness
40 mm
Breast Composition
Breast(50%)
Breast Diameter
17 cm
Lesion Type
Ca5(P04)OH
Lesion thickness
0.1 mm
Lesion Depth
0 mm
Lesion Size
0.1 mm
Task
Equal weighted
Apply Image Task

MTF
NPS
Mean Breast Dose
50 mGy
% Glandular
50
SPR
0
Slot Scan
0
Anti Scatter Grid
JPL GRID-2100
Primary T
0.75
Scatter T
0.36
DQE(f)
0.56
SNR
0
DQE(l)
0.56
SNR
0

Figure A.2: Slot scanning mammography system.

117

A.4.1 Generation of x-ray energy spectrum

In this stage the incident x-ray spectrum of slot scanning mammography is generated using a raw spectrum taken from published data by Boone (1997) and the technical specifications of the Varian x-ray tube. The x-ray spectrum is generated in following steps.

- Select of the focal spot using the pop-up menu *Focal spot* and *Focal spot size*.
- Define the tube inherent filtration using pop-up menu *Inherent filtration*.
- Select the anode material using the pop-up menu *Anode material*.
- Set the technique factors such as tube voltage, tube current, scanning speed, slot width, air gap, source-detector and source-collimator distances.

The *Generate spectrum* button allows the user to generate the x-ray spectrum according to the previous settings, using one of the following m-file.

- *HighTunSpectrum.m*, is used to generated high tungsten energy spectrum (30 - 150keV) using Boone (1997) fitting coefficients.
- *TunSpectrum.m*, is used to generate a low tungsten energy spectrum (20 - 45keV) using Boone (1997) fitting coefficients.
- *MolySpectrum.m*, is used to generate a Molybdenum energy spectrum (20 - 45keV) using Boone (1997) fitting coefficients.

The coefficients of the Boone polynomial equations were saved in text files. The linear attenuation coefficients for different materials were saved in MATLAB data files (MAT-file) and text files. The following are some of these files:

- *TunCoff.txt*, contains the coefficients of the Boone polynomial equation for the tungsten high energy spectrum.
- *TunCoff3.txt*, file contains the coefficients of the Boone polynomial equation for the tungsten low energy spectrum.

- *MolyCoff.txt*, contains the coefficients of the Boone polynomial equation for the Molybdenum energy spectrum.
- *Al1*, this MAT-file contains the linear attenuation coefficients of aluminium for the tungsten high energy spectrum.
- *Al*, this MAT-file contains the linear attenuation coefficients of aluminium for the tungsten low energy spectrum.
- *beryllium.txt*, contains the linear attenuation coefficients of beryllium for the tungsten high energy spectrum.

The *Apply filter* button allows the user to modify the x-ray spectrum using an external filter using M-file *Compounds-Attenuation.m*.

The optional button **Spectrum calibration** allows the user to calibrate the raw spectrum using measured entrance dose at a given distance from the source.

A.4.2 Imaging Task

This section allows the user to define a certain imaging task and calculate the modified energy spectrum incident on the detector. The GUI allows the user to select imaging task parameters such as the thickness and the composition of the breast, and the microcalcification type, size and its position within the breast tissue. The following pop-up, edit menu, and button are used to select a certain image task: *Breast thickness*, *Breast composition*, *Lesion type*, *Lesion depth*, and *Lesion size*.

Task pop-up menu. This allows the user to select a task function such as spatial-frequencies equally weighted, low-frequency detection, and high-frequency discrimination as defined by Siewerdsen and Jaffray (2000).

The linear attenuation coefficients for different breast composition and different microcalcification composition are saved in MATLAB data files (e.g. *0Breatn.MAT*, *50Breatn.MAT*, *100Breatn.MAT*).

The *Apply Image Task* button allows the user to apply an image task according to the previous settings.

A.4.3 Detector Parameters

This section allows the user to input the detector parameters such as scintillator material, thickness, decay time constant, Swank factor, detector coupling, and charge coupled device (CCD) parameters and to calculate the system gain, which include the quantum efficiency of the detector, detector coupling efficiency, and the conversion gain. The user is allowed to test different detector technologies to be used for slot scanning mammography.

The following GUI buttons and the associated the m-files are used for calculation of the detector parameters. The algorithms in Chapter 2, Section 2.3.2.1 were implemented as MATLAB codes.

- The *Quantum efficiency* button and the M-file *Qeff.m* are used to calculate the quantum efficiency of the detector under evaluation.
- The *Conversion gain* button and the M-file *conversion - gain - Csl.m* are used to calculate the conversion gain of the detector.
- The *FOT parameters* button opens the FOT transmission efficiency GUI (*Transm-Eff - FOT.m*) window (Figure A.3) which allows the user to enter the parameters of the fibre optic taper (FOT) and then calculate the transmission efficiency of FOT. The FOT parameters are the refractive index, transmission factor, demagnification ratio, and fill factor. The transmission efficiency of FOT is calculated using the following equation Hejazi and Trauernicht (1997):

$$T_e = \left(\frac{1}{m}\right)^2 \times \left(\frac{\sqrt{n_2^2 - n_3^2}}{n_1}\right)^2 \times T_F \times F_C \quad (\text{A.1})$$

m is demagnification factor; n_1 is refractive index of the air; n_2 is refractive index of core; n_3 is refractive index of cladding; T_F is transmission efficiency; F_C is Fill factor.

- The button *Trans Eff FOT* is used to display the calculated transmission efficiency of FOT.
- The button *Coupling efficiency* allows the user to calculate the total coupling efficiency of the fibre optics (fibre optic plate and the fibre optic taper).

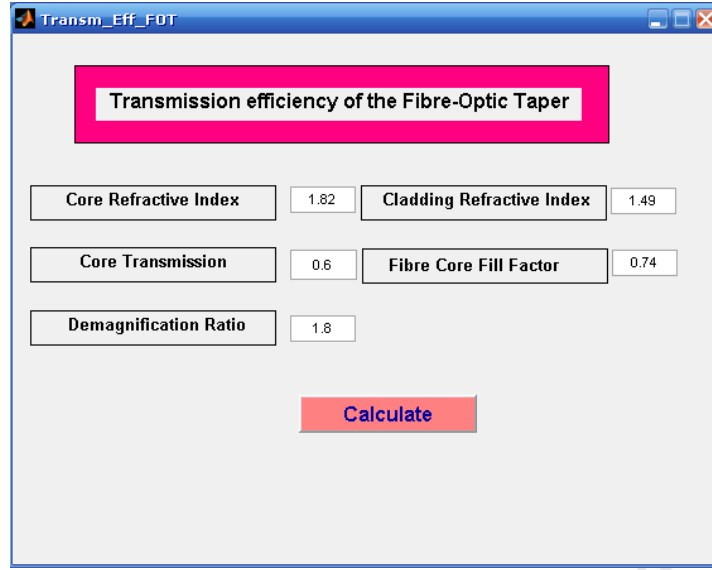


Figure A.3: Transmission efficiency of the fiber optic taper FOT.

- The button *Additive noise parameters* opens the additive noise GUI (*Additive – Noise.m*) (Figure A.4) which allows the user to calculate the total additive noise and save the result in a text file. The input parameters for this GUI are: readout noise, dark current, integration time, pixel size, and demagnification.
- The button *Additive noise* is used to display the calculated additive noise. The dark noise is calculated from the dark current using the following equation (Hejazi and Trauernicht, 1997):

$$D_s = \frac{D_C \times a_{pixel} \times N_{row}}{S_{Speed}} \quad (A.2)$$

D_C is dark current; a_{pixel} is the pixel size; N_{row} is number of TDI row; S_{Speed} is scanning speed.

A.4.4 Output parameters

After the energy x-ray spectrum has been generated and modified by the selected imaging task, and the detector output parameters and the additive noise have been calculated, this section can be used to estimate different beam quality and image quality parameters. The beam quality and the image quality output parameters were either coded within the callback GUI functions or using a separate M-File in the callback GUI function. The system output parameters and the associated M-files include the following:

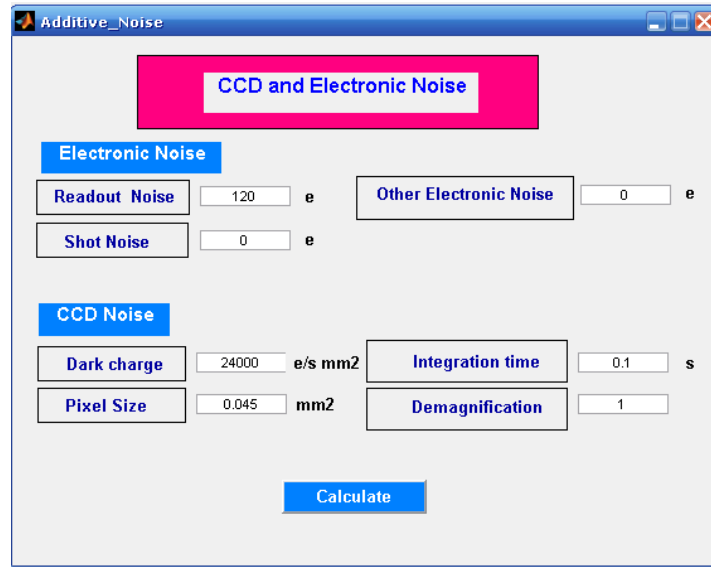


Figure A.4: Window for the calculation of the detector additive noise.

- The *Average Energy*, *HVL*, and *MeanFluence* buttons are used to calculate the mean energy, half value layer, and the photon fluence respectively.
- The *MTF* pop-up menu this allows the user to estimate the system MTF which describes the overall blurring of the system and is the product of the MTF of all the individual stages. The individual stage MTF is calculated using the algorithms described in Chapter 2, Section 3.15 and saved in a text file. The M-files involved in the calculation of the MTF are as follows:
 1. *CsIschrMTF.m* is used to estimate the spatial spreading of optical quanta in the scintillator.
 2. *pixelap.m* is used to estimate the integration of quanta by the CCD characterized by the pre-sampling pixel MTF.
 3. *Gfocalspot.m* is used to estimate the geometrical blurring characterized by the focal spot MTF.
 4. *sintdelay.m* is used to estimate the phosphor persistence blurring characterized by scintillator time decay MTF.
 5. *VmissmatMTF.m* is used to estimate spatial degradation due to the mismatch between the scanning velocity and the charge transfer rate characterized by velocity mismatch MTF.

The *System NPS* pop-up menu allows the user to estimate the system noise power spectrum (NPS) using Equation ??.

The *Calculate DgN* button allows the user to estimate breast dose characterized by normalized glandular dose (NgD) for different breast thicknesses and different breast composition, using the M-file *DNMoly11.m*.

The *SPR* pop-up menu allows the user to estimate the influence of x-ray scatter characterized by the scatter-primary ratio (SPR) for full-field and slot scanning digital mammography using Equation 3.9 and 3.11 and the M-files *calculate-SPR.m* and *calculate-SPR2.m*, respectively. The coefficients of the fitted equations derived by the author (Equation 3.11) using data from Jing *et al.* (1998), and Boone *et al.* (2000), are shown in Table A.1.

Table A.1: The coefficients of the fitted equations used for calculation of SPR for slot scanning mammography

Breast thickness	The fitting coefficients			
	a	b	c	d
2cm	0.0000005	-0.0001	0.0074	0.0096
4cm	0.0000006	-0.0002	0.0216	0.0144
6cm	0.0000020	-0.0003	0.0178	0.0190
8cm	0.0000030	-0.0004	0.0230	0.0216

The *Lesion SNR* and *DQE* buttons allow the user to estimate the signal-to-noise ratio (SNR), zero-frequency and frequency dependent detective quantum efficiency (DQE (0), DQE (f)) using Equation 2.33 and 2.35. The DQE algorithm is implemented as M-file code *Det-Qau-Efficiency.m*.

The *Help* button opens GUI help window.

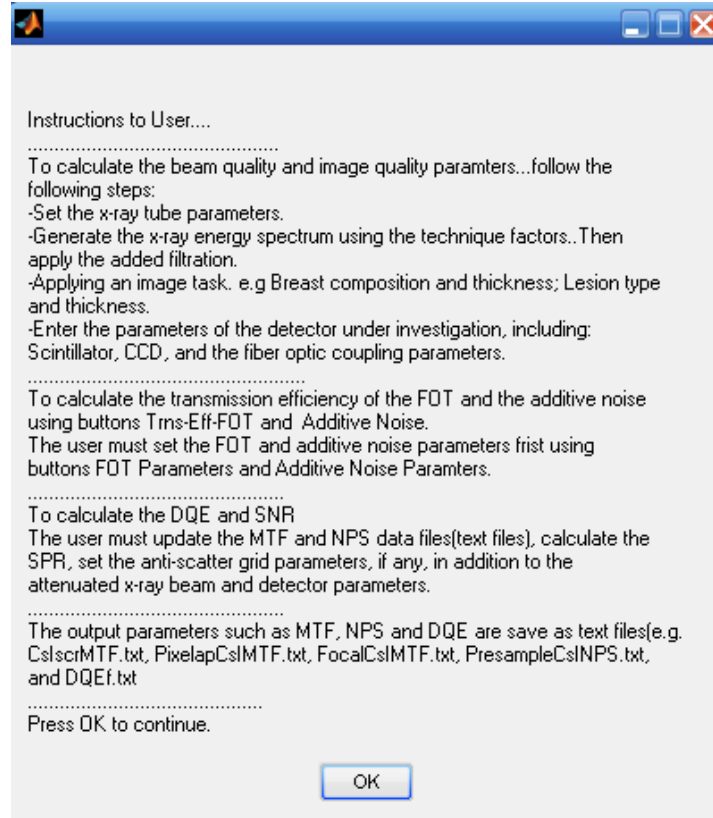


Figure A.5: MammoModel Help window.

A.5 PhantomUser GUI

A MATLAB GUI titled *PhantomUser* was created for optimization of slot scanning digital mammography using a generated mathematical phantom and optimization of the x-ray beam using a monochromatic x-ray filter. The mathematical phantom was generated using the algorithms described in Chapter 5. This model was also used to study the effect of the MXF monochromatic beam compared to the polychromatic beam on detector performance. The image quality parameter algorithms described in Chapter 2 such as image contrast, DQE, CNR and breast dose are used for detector optimization and are implemented as functions in GUI. Figure A.9 shows the GUI for the *PhantomUser*.

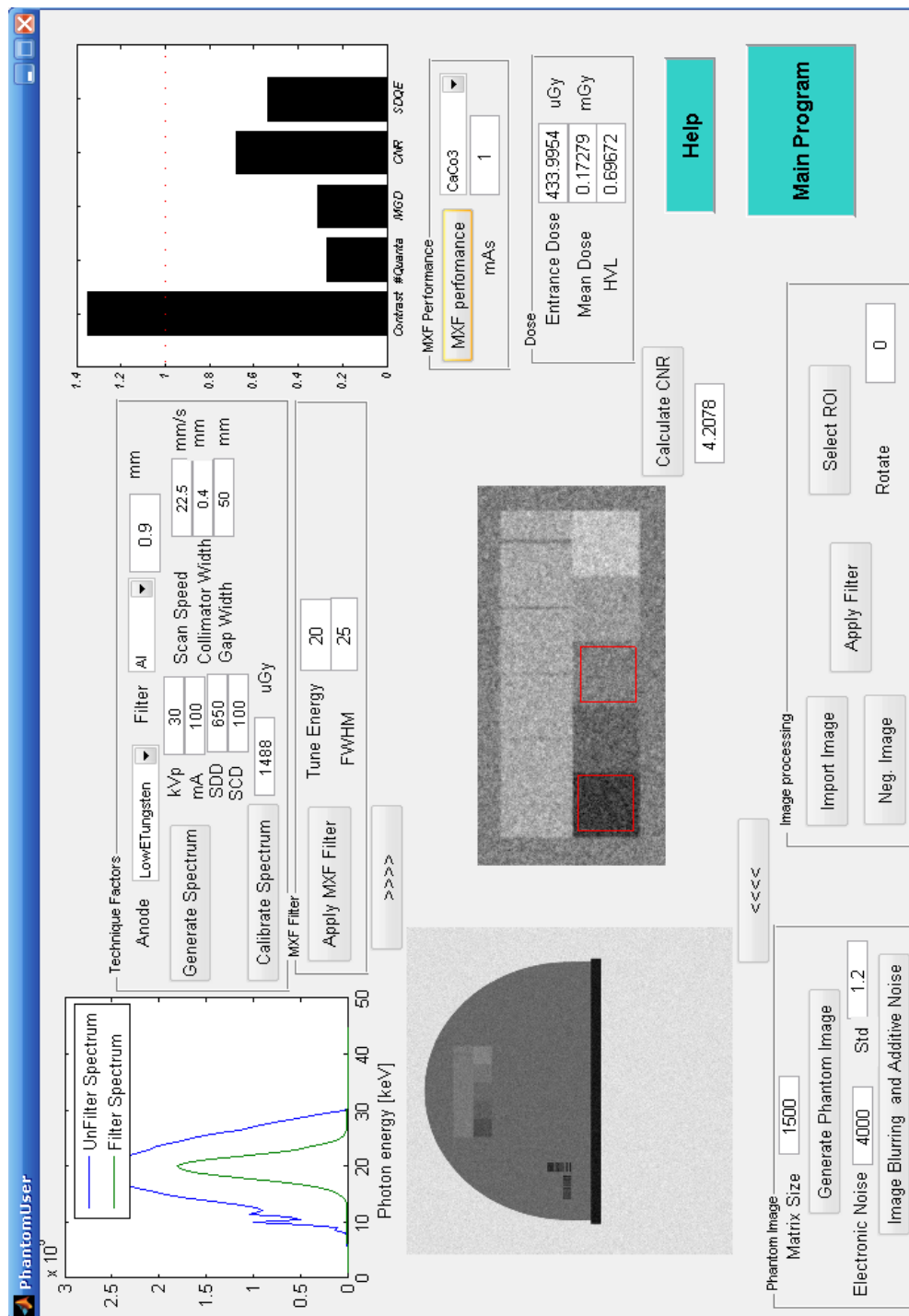


Figure A.6: Multispectrum optimization.

The following steps allow the user to generate the mathematical phantom using the a polychromatic or monochromatic beam:

- Generate the x-ray spectrum for the polychromatic beam using the technique factors or the calibrated entrance dose.
- The *Apply MXf Filter* button allows the user to generate the monochromatic beam using a multispectrum filter of specific tuned energy and FWHM (beam width), as discussed in Chapter 6, Section 6.3.2.
- Select the phantom matrix size (e.g. 1500x1500).
- The *Generate phantom image* button, allows the user to generate an intensity phantom image of different matrix sizes using the calculated energy spectrum . The algorithms used for calculation of the pixel intensity of the phantom body and the structures within the phantom, which are discussed in Chapter 5, were coded as M-files and a callback function in the *phantomUser* GUI. The quantum noise is added at this stage using the M-file *imnoise02.m*.
- Using the *Image Blurring and Additive Noise* button the estimated quantum noise is added to phantom image using the M-file *imnoise03.m*, the Gaussian distortion is applied using the Matlab function *imfilter*, which convolves the Gaussian PSF with the phantom image. Finally, the additive noise is added to the phantom image using the M-file *imnoise02.m*.

A.5.1 Image Processing

This section allows the user to apply some image processing and calculate the contrast-to-noise ratio (CNR). The image processing available is the following:

- The *Select ROI* button is used to select a region of interest in the phantom image with an optional degree of rotation.
- The *Apply Filter* button is used to apply a Wiener filter to the noisy phantom images using the Matlab function *wiener2*.
- The *Neg Image* button, allows the user to invert the generated phantom raw images to resemble actual radiographic images using the M-file *neg-image.m*

- The *Calculate CNR* button allows the user to calculate the contrast-to-noise ratio using Equation 5.6, for the selected region of interest of a target and background area. The regions of interest have an area of 100x100 pixels.
- The *Import image* button allows the user to load the digital images of the physical tissue-equivalent phantom acquired using the mammography system.

A.5.2 MXF Performance

This section allows the user to study the performance of the monochromatic beam using the ratio of image quality parameters and breast dose calculated for the monochromatic and the polychromatic beam. The image quality parameters are, image contrast, breast dose, CNR, and DQE. The following M-files are used in calculation of image parameters; the methods of calculation are discussed in Chapter 2.

- *HighCsIcontrast3.m* is used to calculate the image contrast for the monochromatic beam and the polychromatic beam.
- *TungExposure.m* is used to calculate the exposure of the monochromatic beam and polychromatic beam.
- *DNMoly1.m* is used to calculate breast dose from the monochromatic beam and the polychromatic beam.
- *Optimum-Scan-DQE1.m* is used to calculate the detective quantum efficiency of the monochromatic beam and the polychromatic beam.

The following steps are used to generate the bar plot, which represents the MXF performance:

- A polychromatic beam is generated using the technique factors or the calibrated dose.
- A monochromatic beam is generated using the multispectrum filter of a defined tuned energy and beam resolution (beam width).
- The *MXF performance* button allows the user to generate a bar plot of image quality ratio using the above mentioned M-files.

- mAs represents the factor by which the scanning time (tube load) increases. $2mAs$ means doubling the scanning time.

The *Help* button opens GUI help window.

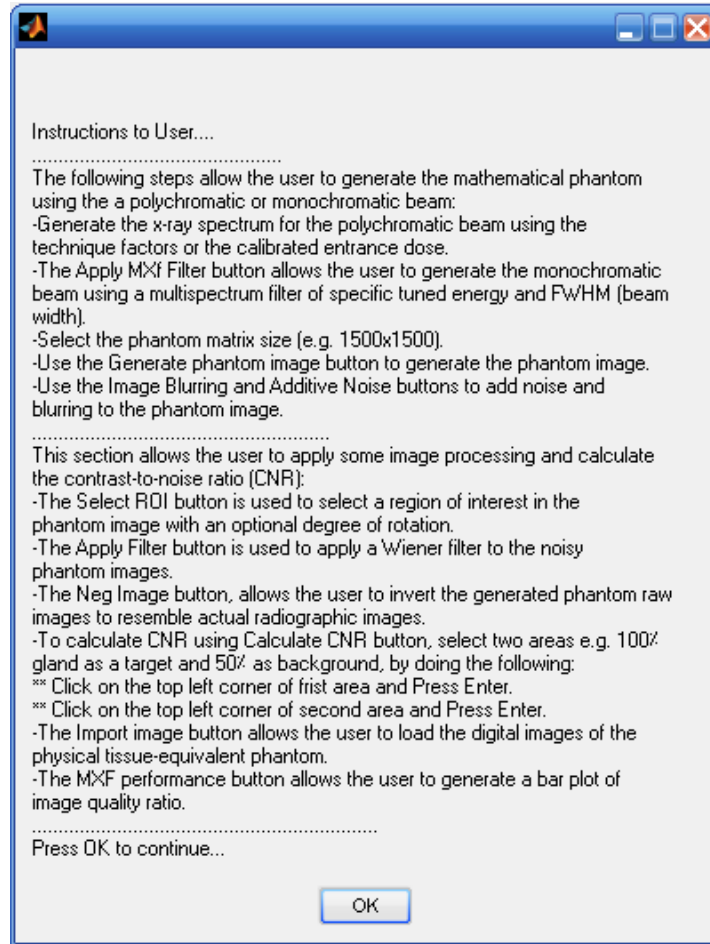


Figure A.7: PhantomUser Help window.

A.6 Optimum Threshold GUI

A MATLAB GUI titled *OptimumThreshold* was created to assist in the implementation of novel algorithms developed by Marchal *et al.* (2006) for the selection of detection thresholds in single-photon counting in digital mammography. Figure A.8 shows the GUI for *OptimumThreshold*. The image quality parameter algorithms described in Chapter 7 are implemented as functions in the GUI.

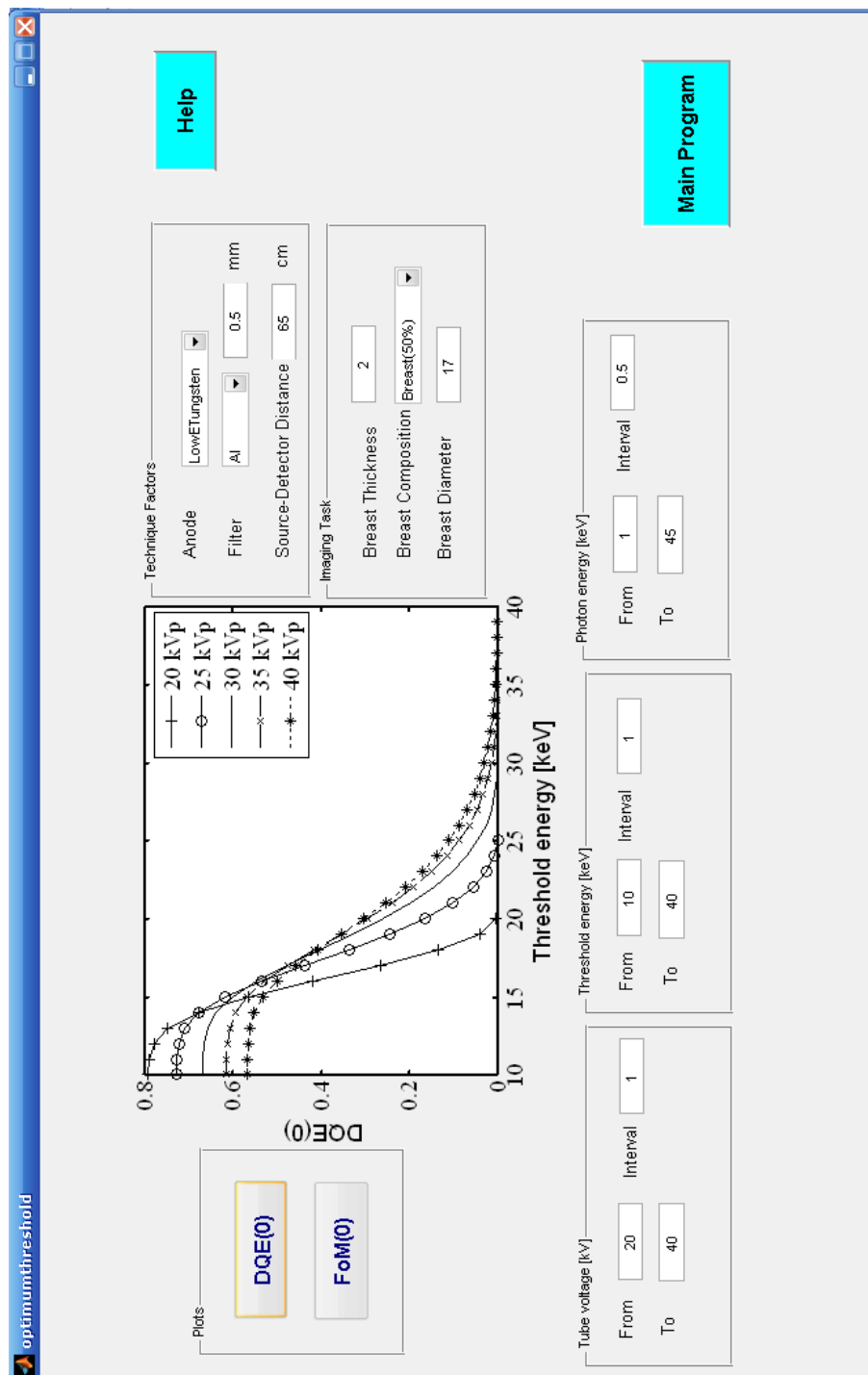


Figure A.8: Photon counting detector optimization.

A.6.1 Technique Factors

This allows the user to generate the energy spectra using the technique factors as discussed in Section A.4.1.

A.6.2 Imaging Task

This allows the user to define an imaging task as discussed in Section A.4.2.

A.6.3 Tube voltage

This allows the user to select a range of different voltages to estimate the effect on the optimum detection threshold energies.

A.6.4 Threshold energy

This allows the user to define the range of optimum detection threshold energies.

A.6.5 Photon energy

This allows the user to define the range of photon energies used to calculate the optimum detection threshold.

The following M-files are involved in calculation of image parameters:

- *DNMoly11.m* is used to calculate mean breast dose.
- *calculate-snr2.m* is used to calculate the broad-spectrum SNR.
- *dqe.m* is used to calculate the broad-spectrum DQE.
- *Diffrr.m* is used to calculate the difference in linear attenuation between the background and the microcalcifications to be detected.

The *Main Program* button allows the user to return to the main menu.

The *Help* button opens GUI help window.

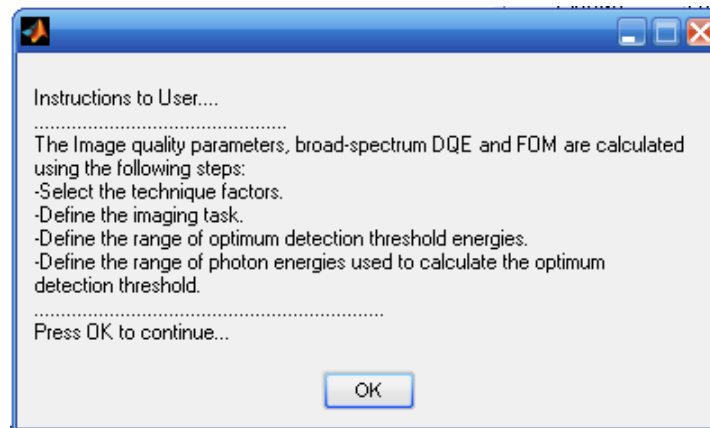


Figure A.9: The OptimumThreshold Help window.

Appendix B

The linear slot scanning mammography system components

The following are some of the components of the linear slot scanning mammography system:

B.1 C-arm and x-ray source



(a)



(b)

Figure B.1: (a) beam shaping collimator; (b) mounted frame

B.2 The collimation system

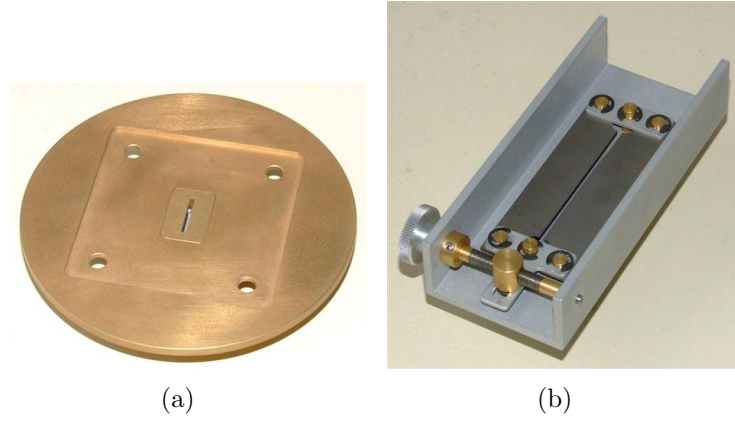


Figure B.2: (a) beam shaping collimator; (b) slot collimator

B.3 The detector and its components

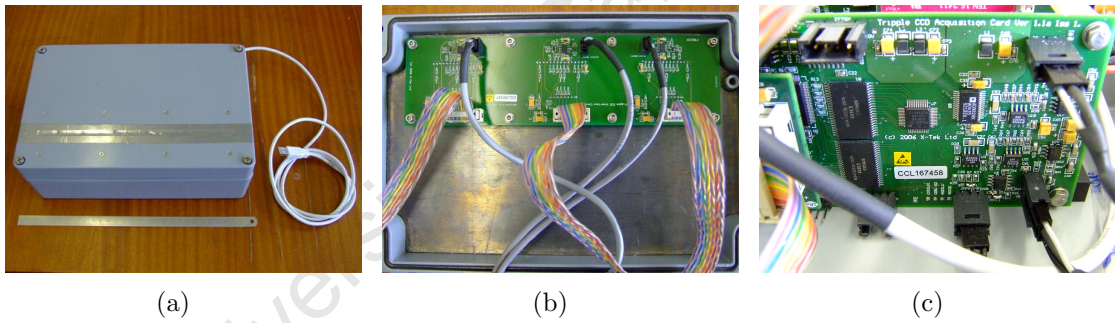


Figure B.3: (a) LSSM detector; (b) Three CCD cameras; (c) CCD acquisition and control card

Appendix C

Hamamatsu charge-coupled device
(CCD) data sheet, April 2003

University of Cape Town

CCD area image sensor S8658-01

Front-illuminated FFT-CCD for X-ray imaging



S8658-01 is a FFT-CCD area image sensor specifically developed for X-ray imaging. Since a FOS (Fiber Optic plate with Scintillator) to convert X-rays into visible light is mounted on the CCD chip, X-ray images can be captured in fine detail. Three CCD chips are linearly arranged in close proximity to form a long and narrow sensor format. Effective size of each active area is 73.728 (H) × 6.144 (V) mm², so the overall active area length of the three chips is up to 220 mm in length. Each CCD chip has 1536 × 128 pixels and the pixel size is 48 × 48 μm.

When used in TDI operation mode which is a special feature of this CCD image sensor, even X-ray images of a moving object can be clearly acquired, making S8658-01 ideal for non-destructive inspection of products carried on a belt conveyor, etc. S8658-01 is pin compatible with S8658. (Operating condition is a little bit changed from S8658)

Features

- FFT-CCD coupled with FOS for X-ray imaging
- 1536 (H) × 128 (V) pixel format
- Pixel size: 48 × 48 μm
- Slit-like image of 220 mm long by aligning 3 CCD chips together
- Coupled with FOS for X-ray imaging
- TDI (Time Delay Integration) operation
- 100 % fill factor
- Wide dynamic range
- Low dark current
- MPP operation

Applications

- General X-ray imaging
- Non-destructive inspection
- Dental panorama, cephalo

■ Selection guide

Type No.	Cooling	Number of total pixels	Number of active pixels	Active area [mm (H) × mm(V)]
S8658-01	Non-cooled	1536 × 128	1536 × 128	73.728 × 6.144

Note) As an input window, FOS is suited to S8658-01.

■ General ratings

Parameter	Specification
CCD structure	Full frame transfer or TDI
X-ray sensitive area	220 × 6 mm
Fill factor	100 %
Number of active pixels	1536 (H) × 128 (V) *1
Pixel size	48 (H) × 48 (V) μm
CCD active area	73.728 (H) × 6.144 (V) mm *1
Vertical clock phase	2 phase and 2 line
Horizontal clock phase	2 phase and 2 line
Output circuit	Two-stage MOSFET source follower with load resistance
X-ray resolution	4 to 6 Lp/mm at 60 kVp, 10 m Roentgen
Reliability	100,000 shots at 60 kVp, 10 m Roentgen
Package	60 pin ceramic package
Window	FOS (Fiber Optic plate with Scintillator)

*1: Number of active pixels per chip. Three chips are used.

PRELIMINARY DATA
Apr. 2003

SOLID
STATE DIVISION

HAMAMATSU

■ Absolute maximum ratings (Ta=25 °C)

Parameter	Symbol	Min.	Typ.	Max.	Unit
Storage temperature	Tstg	-20	-	+70	°C
Operating temperature	Topr	0	-	+40	°C
OD voltage	V _{OD}	-0.5	-	+20	V
RD voltage	V _{RD}	-0.5	-	+18	V
ISV voltage	V _{ISV}	-0.5	-	+18	V
IGV voltage	V _{IGV}	-15	-	+15	V
IGH voltage	V _{IGH}	-15	-	+15	V
SG voltage	V _{SG}	-15	-	+15	V
OG voltage	V _{OG}	-15	-	+15	V
RG voltage	V _{RG}	-15	-	+15	V
TG voltage	V _{TG}	-15	-	+15	V
Vertical clock voltage	VP1AV, VP2AV VP1BV, VP2BV	-15	-	+15	V
Horizontal clock voltage	VP1AH, VP2AH VP1BH, VP2BH	-15	-	+15	V

■ Operating conditions (MPP mode, Ta=25 °C)

Parameter		Symbol	Min.	Typ.	Max.	Unit
Output transistor drain voltage		V _{OD}	12	15	18	V
Reset drain voltage		V _{RD}	12	13	14	V
Output gate voltage		V _{OG}	-0.5	2	5	V
Output transistor ground voltage		V _{SSA}	-	0	-	V
Substrate voltage		V _{SSD}	-5	0	-	V
Test point	Vertical input source	V _{ISV}	-	V _{RD}	-	V
	Vertical input gate	V _{IGV}	-8	0	-	
	Horizontal input gate	V _{IGH}	-8	0	-	
Vertical shift register clock voltage	High	V _{P1AVH, P2AVH} V _{P1BVH, P2BVH}	0	3	6	V
	Low	V _{P1AVL, P2AVL} V _{P1BVL, P2BVL}	-9	-8	-7	
Horizontal shift register clock voltage	High	V _{P1AHH, P2AHH} V _{P1BHH, P2BHH}	0	3	6	V
	Low	V _{P1AHL, P2AHL} V _{P1BHL, P2BHL}	-9	-8	-7	
Summing gate voltage	High	V _{SGH}	0	3	6	V
	Low	V _{SGL}	-9	-8	-7	
Reset gate voltage	High	V _{RGH}	0	3	6	V
	Low	V _{RGL}	-9	-8	-7	
Transfer gate voltage	High	V _{TGH}	0	3	6	V
	Low	V _{TGL}	-9	-8	-7	

■ Electrical characteristics (Ta=25 °C)

Parameter	Symbol	Remark	Min.	Typ.	Max.	Unit
Signal output frequency	f _c		-	2	4	MHz
Reset clock frequency	f _{rg}		-	2	4	MHz
Vertical shift register capacitance	CP1AV, CP2AV CP1BV, CP2BV		-	15000	-	pF
Horizontal shift register capacitance	CP1AH, CP2AH CP1BH, CP2BH		-	500	-	pF
Summing gate capacitance	C _{SG}		-	15	-	pF
Reset gate capacitance	C _{RG}		-	10	-	pF
Transfer gate capacitance	C _{TG}		-	500	-	pF
Transfer efficiency	CTE	*2	0.99995	0.99999	-	
DC output level	V _{out}	*3	5	8	11	V
Output impedance	Z _o	*3	-	500	-	Ω
Power dissipation	P	*3, *4	-	60	-	mW

*2: Measured at half of the full well capacity. CTE is defined per pixel.

*3: V_{OD}=15 V.

*4: Power dissipation of the on-chip amplifier (each chip).

■ Electrical and optical characteristics (Ta=25 °C, unless otherwise noted)

Parameter	Symbol	Remark	Min.	Typ.	Max.	Unit
Saturation output voltage	Vsat		-	Fw × Sv	-	V
Full well capacity	Vertical	Fw	600	1200	-	ke ⁻
	Horizontal		600	1200	-	
	Summing		600	1200	-	
CCD node sensitivity	Sv	*5	0.45	0.6	-	μV/e ⁻
Dark current (MPP mode)	DS	*6	-	8	24	ke ⁻ /pixel/s
Readout noise	Nr	*7	-	60	120	e ⁻ rms
Dynamic range	DR	*8	5000	20000	-	
X-ray response non-uniformity	XRNU	*9, *10	-	±10	±30	%
Blemish	Point defects *11	-	-	-	10	-
	White spots		-	-	10	
	Black spots		-	-	0	
	Cluster defects		-	-	0	
	Column defects	*13	-	-	0	
X-ray resolution	ΔR		4	6	-	Lp/mm

*5: V_{OD}=15 V.

*6: Dark current doubles for every 5 to 7 °C.

*7: -40 °C, operating frequency is 2 MHz.

*8: Dynamic range = Full well capacity / Readout noise

*9: X-ray irradiation of 60 kVp, measured at half of the full well capacity.

*10: XRNU (%) = Noise / Signal × 100

Noise: Fixed pattern noise (peak to peak)

Measuring region that is within 220.0 mm (H) × 6.0 mm (V) (refer to dimensional outline)

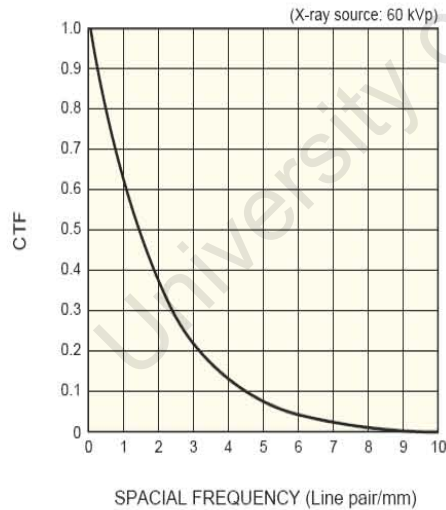
*11: White spots > 20 times of typ. dark signal (8 ke⁻/pixel/s).

Black spots > 50 % reduction in response relative to adjacent pixels, measured at half of the full well capacity.

*12: continuous 2 to 9 point defects.

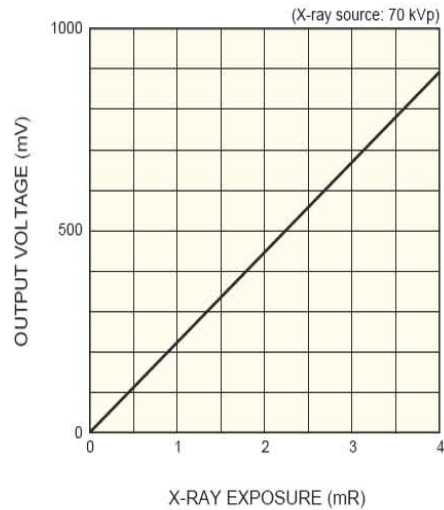
*13: continuous >10 point defects.

■ Resolution



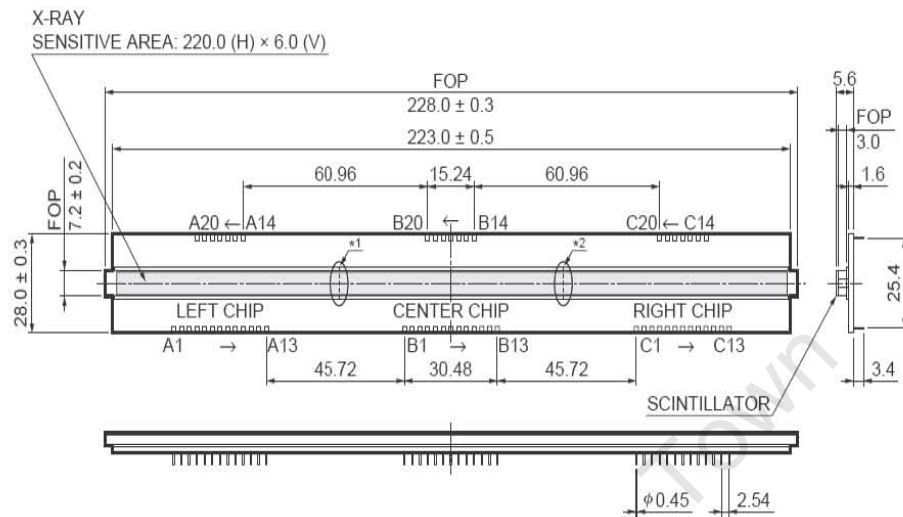
KMP080348EA

■ Response



KMP080348EA

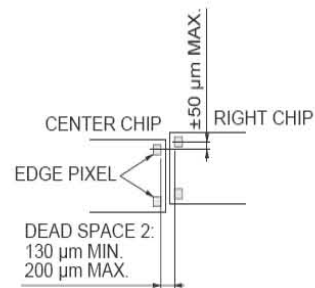
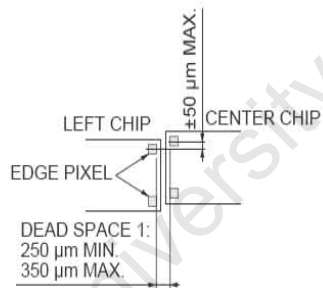
■ Dimensional outline (unit: mm)



KMPD0148EC

*1 Details

*2 Details



Appendix D

Data sheet: Tissue-Equivalent Phantom for Mammography

University of Cape Town

Tissue-Equivalent Phantom for Mammography

*A Refined Quality Assurance Tool
for Today's Advanced Imaging Systems*

Proven simulation technology enables the use of tissue-equivalent, realistically-shaped phantoms for mammographic quality assurance.

CIRS resin material mimics the photon attenuation coefficients of a range of breast tissues. Average elemental composition of the human breast being mimicked is based on the individual elemental composition of adipose and glandular tissue reported by Hammerstein.

Attenuation coefficients are calculated by using the "mixture rule" and the Photon Mass Attenuation and Energy Absorption Coefficient Table of J.H. Hubbell.

The CIRS Model 011A Breast Phantom contains targets that are engineered to test the threshold of the new generation of mammography machines. Model 011A is 4.5 cm thick and simulates an average glandular tissue composition.



Model 011A

The Model 011A was designed to test the performance of any mammographic system. Objects within the phantom simulate calcifications, fibrous calcifications in ducts and tumor masses. Test objects within the phantom range in size from those that should be visible on any system to objects that will be difficult to resolve on the best mammographic systems.

CIRS mammography phantoms are also manufactured in 4 cm, 5 cm and 6 cm thicknesses with various glandular equivalencies.

The methodology and design of these phantoms was developed by Dr. Panos Fatouros and his associates at the Medical College of Virginia.

*Tissue Simulation &
Phantom Technology*

CIRS

2428 Alameda Avenue • Suite 212 • Norfolk, Virginia 23513 • USA
(800) 617-1177 • (757) 855-2765 • FAX (757) 857-0523
www.cirsinc.com • admin@cirsinc.com

Model 011A Specifications

- **Line pair target**
1. 20 lp/mm

- **Ca CO₃ specs**
grain size (mm)
2. .130
3. .165
4. .196
5. .230
6. .275
7. .400
8. .230
9. .196
10. .165
11. .230
12. .196
13. .165

- **Step Wedge**
1 cm thick
14. 100% gland
15. 70% gland
16. 50% gland
17. 30% gland
18. 100% adipose

- **Nylon Fibers**
diameter size (mm)
19. 1.25
20. 0.83
21. 0.71
22. 0.53
23. 0.30

- **Hemispheric Masses**
75% glandular/ 25%
adipose, thickness (mm)
24. 4.76
25. 3.16
26. 2.38
27. 1.98
28. 1.59
29. 1.19
30. 0.90

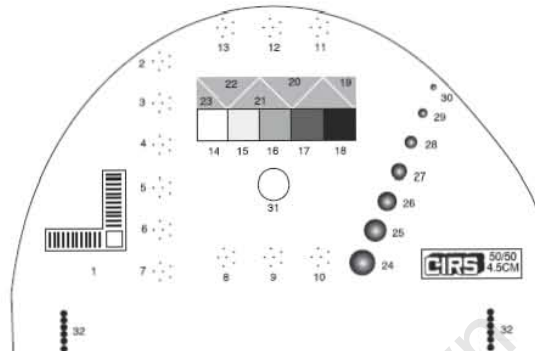
- **Optical Density**
31. reference zone
- **Edge of Beam**
32. localization target
- **Phantom Body**
Length 12.5 cm
Width 18.5 cm
Height 4.5 cm
Material Epoxy

- **Also Included**
30x handheld microscope

Mammography QA documents for recording image evaluations and scores

Technical manual

Carrying case sold separately.



References:

1. Skubic S.E., Fatouros PP. Absorbed Breast Dose: Dependence on Radiographic Modality and Technique, and Breast Thickness. RADIOLOGY, 1986, 161:263-270.
2. Fatouros PP, Skubic S.E., Goodman H. The Development and Use of Realistically Shaped, Tissue- equivalent Phantoms for Assessing the Mammographic Process. RADIOLOGY, 1985 157(p):32.

Appendix E

Matlab Codes

This CD-ROM contains the Matlab code of the mathematical models described in the thesis.

University of Cape Town

References

- Andrew, D. ., Maidment, Yaffe, M. J., Plewes, B. D., Mawdsley, G. E., Soutar, I. C., Starlcoslci, B. G., 1993. Imaging performance of a prototype scanned-slot digital mammography system. *Proceedings of SPIE* 1896, 93–103.
- Aslund, M., Cederstrm, B., Lundqvist, M., Danielsson, M., 2007. Physical characterization of a scanning photon counting digital mammography system based on si-strip detectors. *Medical physics* 34 (6), 1918–1925.
- Bakic, P. R., Albert, M., Brzakovic, D., Maidment, A. D. A., 2002. Mammogram synthesis using a 3D simulation. I. breast tissue model and image acquisition simulation. *Medical Physics* 29 (9), 2131–2139.
- Barbe, D., 1976. Time delay and integration image sensors in solid state imaging. *Proceedings of the NATO Advanced Study Institute on Solid State Imaging (Noordhoff-Leyden)* , 659–671.
- Barnes, G. T., Brezovich, I. A., Witten, D. M., 1977. Scanning multiple slit assembly: a practical and efficient device to reduce scatter. *American Journal of Roentgenology* 129, 497–501.
- Barnes, G. T., Hendrick, R. E., 1994. Mammography accreditation and equipment performance. *Radiographics* 14, 129–38.
- Barnes, G. T., Moreland, R. F., Yester, M. V., Witten, D. M., 1980. The scanning grid: a novel and effective bucky movement. *Radiology* 135, 765–767.
- Barnes, G. T., Wu, X., Sanders, P. C., 1994. Scanning slit chest radiography: A practical and efficient scatter control design. *Radiology* 190, 525–528.

- Beningfield, S., Potogieter, H., Nicol, A., Van, S., Bowie, G., Hering, E., Latti, E., 2003. Report on a new type of trauma full body digital x-ray machine. *Emergency Radiology* 10, 23–29.
- Berger, M., Hubbell, J., Seltzer, S., Coursey, J. S., Zucker, D. S., 2007. XCOM: Photon cross sections database. Last accessed 12 December 2007.
URL <http://physics.nist.gov/PhysRefData/Xcom/Text/XCOM.html>
- Bliznakova, K., Bliznakov, Z., Bravou, V., Kolitsi, Z., Pallikarakis, N., 2003. A three-dimensional breast software phantom for mammography simulation. *Physics in Medicine and Biology* 21 (22), 3699–3719.
- Boone, J. M., 1997. Molybdenum, rhodium, and tungsten anode spectral models using interpolating polynomials with application to mammography. *Medical Physics* 24 (12), 1863–1874.
- Boone, J. M., 2000. *Handbook of Medical Imaging, Volume 1*. Chapter1: X-ray Production, Interaction and Detection in Diagnostic Imaging. Editors J Beutel, H. L. Kundel, R. L. Van Metter, SPIE Press, 2000.
- Boone, J. M., 2002. Normalized glandular dose DgN coefficients for arbitrary x-ray spectra in mammography: Computer-fit values of monte carlo derived data. *Medical Physics* 29 (5), 869–875.
- Boone, J. M., Lindfors, K. K., Cooper, N. I. I., Seibert, J. A., 2000. Scatter/primary in mammography: Comprehensive results. *Medical Physics* 27, 2408–2416.
- Boone, J. M., Siebert, J. A., 1994. A comparison of mono- and polyenergetic x-ray beam performance for radiographic and fluoroscopic images. *Medical Physics* 21, 1853–63.
- Boyle, W. S., Smith, G. E., 1970. Charge-coupled semiconductor devices. *Bell Syst Tech J* 49 (3), 587.
- BreastCancer, 2007. Breastcancer webpage. Last accessed 10 October 2007.
URL <http://www.breastcancer.org>
- Burattini, E., Cossu, E., Maggio, C. D., Gambaccini, M., Indovina, P. L., Marziani, M., Poce, K. M., Simeoni, S., Simonetti, G., 1995. Mammography with synchrotron radiation. *Radiology* 159, 239–244.
- Bushberg, R., 2000. *The Essential Physics of medical Imaging*. McGraw- Hill, New York.

- Cahn, N., Cederstroem, B., Danielsson, M., Hall, A., Lundqvist, M., Nygren, D., 1999. Detective quantum efficiency dependence on x-ray energy weighting in mammography. *Medical Physics* 26, 2680–2683.
- CCDS8658, 2004. Charge couple device (CCD) area image sensor. *HAMAMATSU photonics KK, Solid State Division* .
- Chakraborty, D. P., Eckert, M. P., 1995. Quantitative versus subjective evaluation of mammography accreditation phantom images. *Medical Physics* 22, 133–143.
- Cowen, A. R., Coleman, N. J., 1990. Design of test objects and phantoms for quality control in mammography physics of diagnostic radiology. *IPSM Report 61* , 30–36.
- Cunningham, I. A., 1994. A spatial-frequency dependent quantum accounting diagram and detective quantum efficiency model of signal and noise propagation in cascaded imaging systems. *Medical Physics* 21 (3), 417–427.
- Cunningham, I. A., Yao, J., Subotic, V., 2002. Cascaded linear-systems models and the DQE of flat-panels imagers: Noise aliasing, secondary quantum noise and reabsorption. *Proceedings of SPIE* 4682, 246–249.
- Dainty, J. C., Shaw, R., 1974. *Chapter7: Image Sciences principles, analysis and evaluation of photographic type imaging processes*. Academic press, London, New York, San Francisco.
- Dance, D. R., Hunt, R. A., Bakic, P. R., Maidment, A. D. A., Sandborg, M., Ullman, G., Carlsson, G. A., 2005. Breast dosimetry using high-resolution voxel phantoms. *Radiation Protection Dosimetry* 114 (1-3), 359–363.
- Destounis, S. V., 2008. Acquisition and workflow issues in digital mammography. *Applied Radiology* 37 (10), 10–16.
- deVilliers, M., deJager, G., 2003. Detective quantum efficiency of the lodox system. *SPIE Medical Imaging* 5030, 955–960.
- Diekmann, F., Diekmann, S., Richter, K., Bick, U., Fischer, T., Lawaczek, R., Press, W. R., Schn, K., Weinmann, H. J., Arkadiev, V., Bjeoumikhov, A., Langhoff, N., Rabe, J., Roth, P., Tilgner, J., Wedell, R., Krumrey, M., Linke, U., Ulm, G., Hamm, B., 2004. Near monochromatic x-rays for digital slot-scan mammography: initial findings. *European Radiology* 14, 1641–1646.

- Dobbins, J. T., 2000. *Handbook of Medical Imaging, Volume 1*. Chapter3: Image Quality Metrics for Digital Systems. Editors J Beutel, H. L. Kundel, R. L. Van Metter, SPIE Press, 2000.
- Dobbins, J. T., Ergun, D. L., Rutz, L., Hinshaw, D. A., Clark, H. B. D. C., 1995. DQE(f) of four generations of computed radiography acquisition devices. *Medical Physics* 22, 1581–1593.
- Doyle, P., Martin, C. J., Gentle, D., 2006. Application of contrast-to-noise ratio in optimizing beam quality for digital chest radiography: comparison of experimental measurements and theoretical simulations. *Physics in Medicine and Biology* 51, 2953–2970.
- Egan, R., 1960. Experience with mammography in a tumor institute. evaluation of 1,000 studies. *American Journal of Roentgenology* 75, 894–900,.
- Ewert, U., Zscherpel, U., 2008. A new computer based concept for digital radiographic reference images. Last accessed 10 March 2008.
URL <http://www.ndt.net/article/ecndt02/268/268.htm>
- Fahrig, R., Yaffe, M. J., 1994. Optimization of spectral shape in digital mammography: Dependence on anode material, breast thickness, and lesion type. *Medical Physics* 21 (9), 1473–1481.
- FOS, 1996. Fiber optic plate with scintillator for digital x-ray imaging, technical information. *Hamamatsu, Hamamatsu Corp Bridgewater, NJ* .
- Gambaccini, M., Guerra, A. D., Taibi, A., Tuffanelli, A., 1997. Quasi-monochromatic x-ray sources for mammography via crystal array. *Proceedings of SPIE* , 154–60.
- Gambaccini, M., Taibi, A., Guerra, A. D., Frontera, F., Marziani, M., 1995. Narrow energy band x-rays via mosaic crystal for mammography application. *Nuclear Instrumentation Methods A* 365, 248–54.
- Gambaccini, M., Tuffanelli, A., Taibi, A., Guerra, A. D., 2001. Spatial resolution measurements in quasi-monochromatic x-rays with mosaic crystal for mammography. *Medical Physics* 28, 412–418.
- Ganguly, A., Rudin, S., Bednarek, D. R., Hoffman, K. R., 2003. Micro-angiography for neurovascular imaging. II. cascade model analysis. *Medical Physics* 30 (11), 3029–3039.

- Geise, R. A., Palchevsky, A., 1996. Composition of mammographic phantom materials. *Radiology* 198, 347–350.
- Giersch, J., Niederlohner, D., Anton, G., 2004. The influence of energy weighting on x-ray imaging quality. *Nucl Instrum Methods Phys Res A* 531, 68–74.
- Gobbi, G., Dixit, M. S., Dubeau, J., Johns, P. C., 1999. Photon-counting radiography with the gas microstrip detector. *Physics in Medicine and Biology* 44, 1317–1335.
- Haus, A., 1987. Recent advances in screen-film mammography. *Radiologic Clinics of North America* 25, 913–927.
- Hejazi, S., Trauernicht, D. P., 1997. System considerations in ccd-based x-ray imaging for digital chest radiography and digital mammography. *Medical Physics* 24, 287–297.
- Hessler, C., Depeursinge, C., Grecescu, M., Pochon, Y., Raimondi, S., Valley, J. F., 1985. Objective assessment of mammography systems. *Radiology* 156, 215–219 and 221–225.
- Huda, W., Slone, R., 1995. *Review of Radiologic Physics*. Lippincott Williams and Wilkins, Philadelphia.
- Hunt, R. A., Dance, D. R., Bakic, P. R., Maidment, A. D. A., Sandborg, M., Ullman, G., Carlsson, G. A., 2005. Calculation of the properties of digital mammograms using a computer simulation. *Radiation Protection Dosimetry* 114 (1-3), 395–363.
- ICRU, 1994. *Medical Imaging-the Assessment of Image Quality*. International Commission on Radiation Units and Measurements, Report No. 54, Bethesda, MD.
- Irving, B. J., Maree, G. J., Hering, E. R., Douglas, T. S., 2008. Radiation dose from a linear slit scanning x-ray machine with full body imaging capabilities. *Radiation Protection Dosimetry* 130 (4), 482–489.
- Jing, Z., Huda, W., Walker, J. K., 1998. Scattered radiation in scanning slot mammography. *Medical Physics* 25, 1111–1117.
- Jing, Z., Huda, W., Walker, J. K., Choi, W. Y., 1996. Detective quantum efficiency of a CsI:Tl scintillator based scanning slot x-ray detector for digital mammography. *Proceedings of SPIE* 2708, 633–644.

- Johnston, R. E., Washburn, D., Pisano, E., 1998. Mammographic phantom studies with synchrotron radiation. *Radiology* , 546–555.
- Junck, K. L., DeAlmeida, A., Rezentes, P., Wu, X., Barnes, G. T., 1996. Scatter and grid performance in mammography. *Medical Physics* 23, 1116.
- Khalifah, K. A., Brindhaban, A., AlArfaj, R., Jassim, O., 2006. Image quality of digital mammography images produced using wet and dry laser imaging systems. *Radiography* 12, 13–19.
- Kopans, D. B., 1986. What is a useful adjunct to mammography? *Radiology* 161, 560–561.
- Kopans, D. B., Meyer, J. E., Sadowsky, N., 1984. Breast imaging. *New England Journal of Medicine* 130, 960–967.
- Kroft, L. J. M., Geleijns, J., Mertens, B. J. A., 2004. Digital slit-scan charge-coupled device radiography versus AMBER and bucky screen-film radiography for detection of simulated nodules and interstitial disease in a chest phantom. *Radiology* 231, 156–163.
- Lundqvist, M., 2003. *Silicon Strip Detectors for Scanned Multi-Slit X-Ray Imaging*. Ph.D., Kungl Tekniska Hogskolan, Stockholm, Sweden.
- Maidment, A. D. A., Fahrig, R., Yaffe, M. J., 1993. Dynamic range requirements in digital mammography. *Medical Physics* 20 (6), 1621–1633.
- Mainprize, J. G., Yaffe, M. J., 1996. The effect of phosphor persistence on image quality in digital x-ray scanning systems. *Proceedings of SPIE Vol 2708* , 85–94.
- Manuilskiy, A., Norlin, B., Nilsson, H. E., Frjdh, C., 2004. Spectroscopy applications for the medipix photon counting x-ray system. *Nuclear Instrumentation Methods* 531, 251–257.
- Marchal, J. P., 2005. Extension of x-ray imaging linear systems analysis to detectors with energy discrimination capability. *Medical Physics* 32, 2717–2728.
- Marchal, J. P., Hussein, K., John, L. R., Vaughan, C. L., 2006. Photon counting threshold optimization in mammography. *Nuclear Instruments and Methods in Physics Research A* 563, 137–141.

- Moeckli, R., Verdun, F. R., Fiedler, S., Pachoud, M., Schnyder, P., Valley, J. F., 2000. Objective comparison of image quality and dose between conventional and synchrotron radiation mammography. *Physics in Medicine and Biology* 45, 3509–3523.
- MXF, 2006. Tunable monochromatic x-ray filter. *Monochromatic X-ray Filter Technologies Inc Univ of Maryland park, USA* .
- Nathalie, F., 2007. Breast cancer survival rate. Last accessed 30 December 2008.
URL <http://www.ezinearticles.com>
- NCI, 2004. National cancer institute webpage. Last accessed 1 November 2007.
URL <http://www.nci.nih.gov/cancertopics/types/breast>
- Neitzel, U., Gunther-Kohfahl, S., Borasi, G., Samei, E., 2004. Determination of the detective quantum efficiency of a digital x-ray detector: Comparison of three evaluations using a common image data set. *Medical Physics* 31 (8), 2205–2211.
- NIEHS, 2008. Environmental factors and breast cancer risk. Last accessed 10 March 2008.
URL <http://www.niehs.nih.gov/health/docs/brca-fs.pdf>
- Nishikawa, R. M., Mawdsley, G. E., Fenster, A., Yaffe, M. J., 1987. Scanned projection digital mammography. *Medical Physics* 14 (5), 717–727.
- Pachoud, M., Lepori, D., Valley, J. F., Verdun, F. R., 2004. A new test phantom with different breast tissue compositions for image quality assessment in conventional and digital mammography. *Physics in Medicine and Biology* 49, 5267–5281.
- Rabbani, M., Shaw, R., Van Metter, R. L., 1987. Detective quantum efficiency of imaging systems with amplifying and scattering mechanisms. *J Opt Soc Am A* 4, 895–901.
- Rabbani, M., Van Metter, R. L., 1989. Analysis of signal and noise propagation for several imaging mechanisms. *J Opt Soc Am* 6, 1156–1164.
- RAD-70, 2004. Data sheet of rad70 x-ray tube.
- Reiser, I., Sidky, E. Y., Nishikawa, R. M., 2006. Development of an analytic breast phantom for quantitative comparison of reconstruction algorithms for digital breast tomosynthesis. *Lecture notes in computer science* 4046, 190–196.
- Rose, A., 1973. Vision: Human and electronic. *New York: Plenum* , 21–30.

- Rowlands, J. A., Hunter, D. M., 1991. X-ray imaging using amorphous selenium: A photo-induced charge readout method for digital mammography. *Medical Physics* 18, 421–431.
- Rowlands, J. A., Yorkston, J., 2000. *Handbook of Medical Imaging*. SPIE, Bellingham, Washington 98227-0010, USA. Editors: J. Beutel AND H. L. Kundel AND R. L. Van Metter.
- Samei, E., Lo, Y. J., Yoshizumi, T. T., Jesneck, J. L., Dobbins, J. T., C. E. Floyd, C. E., McAdams, H. P., Ravin, C. E., 2005. Comparative scatter and dose performance of slot-scan and full-field digital chest radiography systems. *Radiology* 235 (3), 940–949.
- Samei, E., Saunders, R. S., Lo, J. Y., Dobbins, J. T., Jesneck, J. L., Floyd, C. E., Ravin, C. E., 2004. Fundamental imaging characteristics of a slit-scan digital chest radiographic system. *Medical Physics* 31, 2687–2698.
- Scheelke, M., 2005. *System characterisation of a full body slit scanning radiography machine: Theory and experiment*. Biomedical engineering, University of Cape Town (UCT), Faculty of Health Sciences, University of Cape Town , Cape Town, South Africa.
- Siewerdsen, J. H., Antonuk, L., 1998. DQE and system optimisation for indirect-detection flat-panel imagers in diagnostic radiology. *Proceedings of SPIE* 3336, 546–555.
- Siewerdsen, J. H., Antonuk, L. E., El-Mohri, Y., Yorkston, J., Huang, W., Cunningham, I. A., 1998. Signal, noise power spectrum, and detective quantum efficiency of indirect-detection flat-panel imagers for diagnostic radiology. *Medical Physics* 24, 614–628.
- Siewerdsen, J. H., Jaffray, D. A., 2000. Optimisation of x-ray imaging geometry (with specific application to flat-panel cone-beam computed tomography. *Medical Physics* 27, 1903–1914.
- Siewerdsen, J. H., Waese, A. M., Moseley, D. J., Richard, S., Jaffray, D. A., 2004. Spektr: A computational tool for x-ray spectral analysis and imaging system optimization. *Medical Physics* 31 (11), 3057–3067.
- Smith, A. D. C., Smith, I. A. C., Dance, D. R., 1998. Objective assessment of phantom image quality in mammography: a feasibility study. *British Journal of Radiology* 71, 48–58.

- Smith, A. P., Hall, P. A., Marcello, D. M., 2004. Emerging technologies in breast cancer detection. *Radiology Management* 26 (4), 16–24.
- Sorenson, J. A., Floch, J., 1985. Scatter rejection by air gaps: an empirical model. *Medical Physics* 12, 308–316.
- Sorenson, J. A., Niklason, L. T., DF, K., 1980. Performance characteristics of improved antiscatter grids. *Medical Physics* 7, 525–528.
- Stierstorfer, K., Spahn, M., 1999. Self-normalising method to measure the detective quantum efficiency of a wide range of x-ray detectors. *Medical Physics* 26 (7), 1312–1319.
- Tapiovaara, M. J., Wagner, R. F., 1985. SNR and DQE analysis of broad spectrum x-ray imaging. *Physics in Medicine and Biology* 30, 519–529.
- Tesic, M. M., Piccaro, M. F., Munnier, B., 1999. Full field digital mammography scanner. *European Journal of Radiology* 31, 2–17.
- Tkaczyk, J. E., LeBLanc, J., Nevin, R., Kautz, G., Albagli, D., Sandrik, J., Granfors, P. R., 2001. Modelling the x-ray energy characteristics of DQE for full-field digital mammography. *AProceedings of SPIE, Vol 4320* , 570–581.
- Veldkamp, W. J., Karssemeijer, N., 1994. Accurate segmentation and contrast measurement of microcalcifications in mammograms: a phantom study. *Medical Physics* 25, 1102–1110.
- Verdun, F. R., Moeckli, R., Valley, J. F., Bochud, F., Hessler, C., 1996. Survey on image quality and dose levels used in europe for mammography. *British Journal of Radiology* 69, 762–768.
- X-Tek, 2006. X-tek systems ltd. Last accessed 10 March 2008.
URL <http://www.xtekxray.com>
- Zhao, W., Ristic, G., Rowlands, J. A., 2004. x-ray imaging performance of structured cesium iodide scintillators. *Medical Physics* 31 (9), 2594–2605.