Optimization of CT Images for Breast Cancer Screening Using Spectral Imaging

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Abstract

About 1.7 million new cases of breast cancer were estimated by the World Health Organization (WHO) in 2012, accounting for 23 percent of all female cancers. In the UK 33 percent of women aged 50 and above were diagnosed in the same year, thus positioning the UK as the 6th highest in breast cancer amongst the European countries. The national Screening programme in the UK has been focused on the procedure of early detection and to improve prognosis by timely intervention to extend the life span of patients. To this end, the National Health Service Breast Screening Programme (NHSBSP) employs 2-D planar mammography because it is considered to be the gold standard technique for early breast cancer detection worldwide. Breast tomosynthesis has shown great promise as an alternative method for removing the intrinsic overlying clutter seen in conventional 2D imaging. However, preliminary work in breast CT has provided a number of compelling aspects that motivates the work featured in this thesis. These advantages include removal of the need to mechanically compress the breast which is a source of screening non-attendances, and that it provides unique cross sectional images that removes almost all the overlying clutter seen in 2D. This renders lesions more visible and hence aids in early detection of malignancy. However work in Breast CT to date has been focused on using scaled down versions of standard clinical CT systems. By contrast, this thesis proposes using a photon counting approach. The work of this thesis focuses on investigating photoncounting detector technology and comparing it to conventional CT in terms of contrast visualization. Results presented from simulation work developed in this thesis has demonstrated the ability of photoncounting detector technology to utilize data in polychromatic beam where contrast are seen to decrease with increasing photon energy and compared to the conventional CT approach which is the standard clinical CT system

Key words: Photoncounting, Conventional CT Lesion, Calcification, Contrast, Mammography.

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Abbreviations and Acronyms

BrCT .......... Breast Computed Tomography
CT .......... Computed Tomography
CDF .......... Contrast Degradation Factor
CAS .......... Computed Aided Software
DBrCT ....... Dedicated Breast Computed Tomography
DBT .......... Digital Breast Tomosynthesis
DCIS .......... Ductal Carcinoma \textit{in situ}
DLA .......... Diffusion Limited Aggregation
FBP .......... Filtered Back Projections
FFDM .......... Full Field Digital Mammography
FOV .......... Field of View
MGD .......... Mean Glandular Dose
MPV .......... Mean Predictive Value
MRI .......... Magnetic Resonance Imaging
NHSBSP ...... National Health Service Breast Screening Programme
NHS .......... National Health Service
PET .......... Positron Emission Tomography
PCBrCT ...... Photon Counting Breast Computed Tomography
PCDs .......... Photon Counting Detectors
RSNA ........ Radiobiological Society of North America
SDD .......... Source to Detector Distance
SFM .......... Screen Film Mamography
SPCT .......... Spectral Computed Tomography
SIC .......... Source to Isocentre
SPECT .......... Single Photon Emission Computed Tomography
US .......... Ultrasonography
WHO .......... World Health Organization
Chapter 1

Introduction

Globally, breast cancer is the most common cancer in women aged 50 years and above, it is known to be the second most common cancer world wide after lung cancer and represents the most frequently diagnosed cancer amongst women in the UK [39],[9],[13],[16]. In 2012, about 1.7 million new cases of breast cancer were estimated by the World Health Organization (WHO)[23]. This number accounted for 23 percent of all female cancers [17]. In the same year 33 percent of women aged 50 and above were diagnosed of breast cancer in the UK. Since the mid 1970s incidence rates in the UK have risen by 7 percent yearly, indicating that 8 out of 10 cases diagnosed are aged 50 and above. This has positioned the UK with the sixth highest incidence rate of breast cancer in Europe[31]. In spite of early diagnosis through screening programmes and modern treatment modalities, breast cancer is still more prevalent albeit with a lower mortality rate in developed countries than it is in the developing countries[15]. This is as a result of different methods of screening, divergent lifestyle and genetic considerations[1].

Clinically, detection of breast cancer with reduced dose and increased sensitivity in women with dense breast has become a major concern as younger pre-menopausal women have begun to undergo screening routinely. This is based on the fact that dense breast are made of more glandular and a greater volume of fibroglandular tissue, which is related to a higher risk of breast cancer[44]. Imaging technologies explored include planar X-ray Mammography, Ultrasonography (US), Magnetic Resonance Imaging (MRI), Scintimammography, Single Photon Emission Computed To-
mography (SPECT), and Positron Emission Tomography[30][5]. Presently 2-D planar X-ray Mammography is widely considered to be the gold standard technique for early breast cancer detection[13], and its use has been employed by the National Health Service Breast Screening Programme (NHSBSP) [9]. Despite its usefulness and high rating, the modality is not without limitations. It suffers the limitations of superposition of 3D structures on to the 2D projected image that can make the presence of breast cancer ambiguous in its appearance [25]. See figure 1.1

![2-D mammography showing breast cancer](image)

Figure 1.1: 2-D mammography showing breast cancer [22].

The emergence of Dedicated Breast Computed Tomography (DBrCT) has shown potential for overcoming the overlapping tissue or superposition of lesion masses and moreover, removes the need for mechanical breast compression which is uncomfortable and in some cases painful[13]. The subject lies prone on the table with a cut out for the breast in a pendate position as shown in figure 1.2. An X-ray tube and a detector placed under the table rotates 360 degrees around the breast from which cross-sectional slices can be reconstructed [18].

This study aims to utilizes computer simulation to investigate the emerging technology
1.1 Breast Screening

1.1.1 Breast Anatomy

The main constituent of the breast include dense and fatty tissues. The dense tissue, made of glandular and connective tissues appears bright on a mammogram, while fatty tissue appears dark[29]. The functional tissue that makes up the breast is responsible for the generation of milk during lactation and size of the breast is dependent on the quantity of fatty tissues[16].

The breast is made up of skin, fat, ducts and supporting fibre tissues as can be seen in figure 1.3. The lobes in the breast are arranged between 15-20 segments, in a circular trend with small structures called lobules [50]. The density of the breast is determined by connective tissues; blood vessels, lymph vessels, fat, glandular tissue and lymph nodes[46].

Figure 1.2: Breast examination with the breast in pendate position[22]
Chapter 1. Introduction

The process of abnormal growth called cancer results in the formation of either malignant or benign tumours. Malignant cancer is the most destructive and may spread to other organs. This is unlike benign tumours that grow gradually and without spreading to other parts of the body[64]. By extension, the anomalous growth of cells that spread by breaking away from the original tumour of breast tissue structure (soft tissue masses) are referred to as metastatic breast cancer[21].

1.1.2 Breast Cancer

Breast cancer may begin in the cells of the lobules or the duct and can be classified to be invasive or non-invasive depending on the diagnosis[21]. The non-invasive cancer referred to as carcinoma \textit{in situ} does not penetrate normal tissues of the breast, whilst,
the invasive cancer does penetrate and affect healthy tissues. About 70-80 percent of breast cancers diagnosed are invasive[65]. Figure 1.4 shows mammography images of breast cancer.

![Mammography images of breast cancer](image)

Figure 1.4: Mammography images of highly visible well differentiated masses showing clear signs of breast cancer[36]

1.1.3 Breast Screening program in the UK

In the Forrest report of 1986 [63], routine screening of women aged 50 and above was recommended in order to increase the life expectancy of women. In line with the report, the National Health Service (NHS) implemented the NHS Breast Screening Programme (NHSBSP) in 1988, using the then gold standard of traditional film screen mammography system in conjunction with self breast examination. Women between the ages of 50-64 years were invited to have mammography examination every three years. Evidently, 95 percent of women feel that accepting breast screening programme is advantageous and about 20 percent relative risk reduction in breast cancer mortality is achieved as a result of the screening programme[21],[38]. Women below the age of 50 years were not initially invited for screening because incidence rate of breast cancer
was lower for that age group[21]. Moreover, these younger pre-menopausal women also tended to have higher levels of glandular tissue present, which limits the sensitivity of conventional 2-D mammography.

In more recent years, film screen mammography has been replaced by 2-D digital mammography. However, the constraint of overlapping dense fibroglandular tissues within the breast decreases the perceptibility of malignant tumours, thereby reducing the specificity of screening with relatively high false -positive recalls[21]. The NHSBSP broadened the screening age spectrum to encompass women with dense breast tissue aged 45-50 years[63]. The independent UK panel on breast cancer screening in 2012 revealed that for every 10,000 UK women aged 50 years and invited for screening, in twenty years time (as at 2012), 43 lives would have been saved[38].

The research into Digital Breast Tomosynthesis (DBT) has been on the increase over a decade with very promising consequences[20]. It is at the moment regarded in clinical imaging as a standard of a future generation X-ray imaging technology because of its 3-D imaging, contrary to the conventional mammography. Although the Planar 2D-X-ray mammography technique have gained much acceptance in clinical breast screening, DBT is being examined as an alternative, due to the intrinsic limitation[55]. DBT as a 3-D imaging modality requires minimal dose images of fixed and condensed breast. The distinctive images are reconstructed into an array of high resolution slices parallel to the detector plane that can subsequently be made visible separately. This technology addresses the symbolic issues of mammography, such as false positive rates and missed cancers. Digital Breast Tomosynthesis is not without limitations as it generates higher X-ray doses and extends reporting time contrary to FFDM[20]. The limitations has being the reason for a growing interest in Computed Tomography CT where true 3-D breast images are produced. More recently, Dedicated Breast Computed Tomography (DBrCT) has shown promise to be a more comfortable technology than mammography, with the potential benefit for sensitivity and specificity[35]. These technologies are considered with a focus on the dose limits administered to patients during screening.
1.1.4 Dose Consideration in Mammography

The effectiveness of mammography in breast cancer detection has been studied\cite{67} where sensitivity of breast tissues to radiation during mammography is seen to be a concern. It is therefore important to monitor the level of radiation dose to patients during mammography screening. This is to retain a relationship between dose and risk to radiation dose where minimal level of dose is used to sustain acceptable rates of breast cancer detection, while the image quality is not compromised. Radiation dose is defined in terms of absorbed energy per unit mass of a tissue expresses in Gray (Gy) which is the international system unit of radiation and is the same as 1 joule per kilogram or 100 rad. It can also be expressed in millisievert defined in terms of average accumulated background radiation dose in a years interval. Typically the average dose used in mammography is 1.5mGy per view in digital mammography.

Radiation energy generated in an x-ray tube passes through the breast and an image is formed. This is usually based on changes in density of tissues in the breast as beam of X-rays passes through the breast to a detector. X-rays photons that fails to penetrate the breast and reach the detector are then absorbed within the breast as radiation dose during mammography screening\cite{35}. Quality assurance in mammography is a strict measure as healthy women are invited for screening, the target is to keep radiation dose as low as possible in line with the principle of "As Low As Reasonably Achieved" (ALARA). Measurement of radiation dose is considered so as to measure the performance of mammography equipment and to compare imaging systems, and also consider dose levels administered to patients.

Dose is most often described as CT dose index, which describes dose to a particular position on an object that is scanned and in slices. An estimate of dose for a given examination therefore depends on on a weighted average of CT dose index\cite{24}. The image quality is described in terms of contrast, spatial resolution, image noise and artifacts. Therefore the usefulness of CT is its capacity to detect points of low contrast during examination. Although this process can be hampered by noise which is related to radiation dose. Radiation dose depends on the tube current, slice scan time and peak kilo-voltage. Where tube current and scan time are considered as mAs relating to
radiation dose and image quality. mAs is therefore directly proportional to radiation dose. Also an increase in peak kilo-voltage increases the intensity of X-rays photons passing through the object to the detector which likely reduces dose as higher peak voltage does not significantly increase patients’ dose[24].

1.1.5 Absorbed Dose

The effect of radiation in a tissue is measured by absorbed dose, which is estimated by finding the quotient of energy deposited in a small amount of volume of tissue or material and its mass. In other words, it is the energy deposited per unit mass of tissue[67]. Absorbed dose is directly related to biological tissue and expressed in Gy. Historically, the unit of absorbed dose has been the rad defined as 1 joule/kilogram, therefore, $1\text{Gy} = 100\text{rad}$ and should be measured when radiation exposure creates a physical or chemical effect in breast which is the absorbing material during mammography screening. In CT it is referred to as Computed Tomography Dose Index CTDI and can be determined at the centre of the slice[19][5], which basically indicates tissue dose. Aside absorbed dose, total radiation to the patient known as the Dose Length Product DLP is considered, which is essentially the sum of radiation through out a CT procedure. It is therefore important to monitor the amount of radiation dose during screening. This is to retain a relationship dose and risk such that dose is considerably reduced while the image quality is not compromised. According to Haus etal(2001) dose level in patients during mammography is a dependable parameter in the evaluation of mammographic process and clinical application.

1.2 Dose Equivalent

In an attempt to discuss sievert $Sv$ which is also a unit of biologically equivalent dose, dose equivalent is also discussed in brief. Deposition of energy in tissue is either in form of heavy charged particles or electrons and hence different magnitude in effect. Damage caused by ionizing radiation is linked to chemical alteration of biological tissues influenced by excitation due to radiation. the extent of damage depends on the level
of energy deposition along a given track called linear energy transfer $L$. Radiation with large $L$ from heavy charged particles has more biological damage compared to electrons with the same energy deposited per unit mass. Dose equivalent therefore quantifies biological effect of a given radiation exposure given by:

$$H = DQN$$

where; $H =$ Dose equivalent, $D$ is absorbed dose with a unit of $J/Kg$, $Q$ is dimensionless quantity factor and $N$ is product of other modified factors. $J/Kg$ has the name Gray when it is absorbed dose and sievert when it is dose equivalent and has a traditional unit of $rem$, $1Sv = 100rem$

### 1.3 Motivation and Aims

Dedicated breast CT has shown promise for clinical investigation to address the limitations of 2-D mammography[21]. However, breast CT to date has only utilized the same approach as used in conventional CT with energy integrating detectors and do not take advantage of the polychromatic energy in the X-ray beam, which has the potential to provide better discrimination of different tissue types[56]. This thesis investigation proposes to examine using spectral imaging methods for breast CT. Spectral imaging is considered because of its distinct advantage of material decomposition over conventional CT[56]. The investigation will be carried out using simulation methods to optimize the use of this approach to breast cancer imaging. This will involve optimizing spectral decomposition methods, imaging acquisition methods, and reconstruction methods in order to validate the approach.

This study aims to utilizes computer simulation to investigate the emerging technology of dedicated breast CT based on photon counting detectors. The number of energy bins for spectral imaging, optimizing geometry for realistic imaging and spectral decomposition will be investigated. This study contributes to knowledge as investigations with photon counting approach can aid the determination of improved contrast and relative dose. More importantly that the binning process is considered during investigations of lesions and calcification in breast imaging.
Chapter 1. Introduction

1.4 Overview of Thesis Structure

To achieve the objectives described in section 1.2, work has been completed to investigate dedicated breast CT based on photon counting detectors with special interest in the energy bins and spectral decomposition for lesion and calcification detection with respect to background in a breast phantom. This thesis is presented as follows;

Imaging technology for breast cancer screening and principles of mammography has been reviewed in chapter 2. The chapter provides information about the key technologies for breast cancer screening, it also provides reasons to investigate the emerging technology for dedicated breast CT based on photon counting detectors. Radiation dose to patients has also been reviewed in this chapter.

Simulation of breast lesions and calcification forms the basis for this thesis and is the starting point for which simulation frame work has been reviewed in chapter 3. The chapter also describes the methodology used in achieving the aim of this thesis, with special consideration to the Siddon Algorithm, phantom designs, simulation process and Filtered Back Projection (FBP).

Chapter 4 provides results based on the findings. The results from simulation work is made of the spectral decomposition for lesion and calcification for mono projections and poly-chromatic projections at different energy bins. An in dept discussion of results is also presented in this chapter.

The conclusion and future work, has been described in chapter 5 with respect to the spectral decomposition and attenuation values of the tissues in the phantom. The conclusion is also described in terms of photon counting approach and conventional CT approach.
Chapter 2

Review of Imaging Technology for Breast Cancer Screening

This chapter discusses the key technologies for breast cancer screening and the motivation to investigate the emerging technology of Dedicated Breast Computed Tomography (DBrCT) based on photon counting detectors. Various imaging technologies are reviewed with Full Field Digital Mammography, considered to be the gold standard for breast screening. Breast cancer screening suggests investigative examination for women with the aim of early detection of disease for better outcome and improvement of life years dependent on the method used [19][5]. Clinically, detecting tumour masses at an early stage has been the primary goal of breast imaging, preferably at less than 10 mm to allow early treatment [30]. The distinctive goals of imaging are: Identification of abnormal tissues, locating the abnormalities within the breast with the lowest radiation dose achieved and to aid decision making process. It is imperative to have sensitivity, specificity and positive predictive value to be higher than 87-90 percent, 58-60 percent and 13-15 percent respectively [21]. Till date the standard imaging approach used in breast screening is the Full Field Digital mammography (FFDM).
Chapter 2. Review of Imaging Technology for Breast Cancer Screening

2.1 Full Field Digital Mammography (FFDM)

Full Field Digital Mammography (FFDM) is currently an imaging gold standard technique used for the early detection of breast cancer [12]. The equipment has a significant improvement in contrast and spatial resolutions which is capable of image manipulation and improved microcalcification detection [49] compared to Screen Film Mammography (SFM). It converts an incident X-ray flux into 2-D mammographic images of the breast. It has replaced the traditional SFM that have served for over 30 years in the UK [12]. Its ability to adjust the contrast, brightness, orientation and direction of images provides better screening results than the SFM. It also reduces examination time, improve visibility, reduce storage space and allows transfer of information easily [12] [61]. Figure 2.1 shows SFM and FFDM images of the breast for comparison.

![Figure 2.1: Distinction between FFDM and SFM][36]

Although FFDM reduces mortality at relatively low cost, 30 percent of all breast cancers are not detected by standard screening, based on the earlier mentioned limitations [13]. These limitations have prompted research into other imaging technologies.
2.2 Digital Breast Tomosynthesis (DBT)

The use of Digital Breast Tomosynthesis (DBT) in x-ray imaging has been explored for over a decade now. The quasi three dimensional imaging technology which is still under investigation has provided useful and promising results for overcoming the limitations in FFDM based on research into its optimization[20]. Schematic shown in figure 2.2. Clinically proven results have demonstrated the increase in sensitivity of DBT compares to FFDM as lesions becomes clearer and improves the accuracy of mammography with clear areas of the overlapping tissue[20] [55]. However, its ability to detect microcalcification is inferior to FFDM.

Breast imaging using DBT was first explained by Niklason etal[42]. The geometry of this technology is very much close to that of FFDM, the exceptional difference is that, the source of x-rays revolves around a compressed and static breast to acquire
finite number of projections. The low dose projections are reconstructed to produce pseudo 3-D images that helps overcome the tissue superposition issue and provides pseudo tomographic views of the breast at different depths[20]. Figure 2.3 shows a clear difference in lesions visibility between DBT and 2-D mammography. In addition to the advantage of minimizing the concealed tissue superposition which minimizes sensitivity and specificity in FFDM, DBT also controls the loss of data in the third dimension. Based on the works of Niklason et al.[42], where they compared the perceptibility of breast cancer in DBT and in FFDM, DBT may be a superior technology to FFDM. In fact, a study presented at the Radiological Society of North America (RSNA) in 2013 shows a university in Philadelphia hospital is already using DBT as a technique for breast cancer screening. However, there is no overall consensus on the manner in which DBT should be used in a screening context[54].

![Figure 2.3: A display of images showing clear distinction of lesion in (a) DBT image and (b)2-D mammographic image](image)

In spite of the advantages it offers, there exist some fluctuations in DBT imaging which adversely interfere with the quality of reconstructed images. This may include the
number of projections during acquisition, reconstruction algorithm, it also increases the amount of scatter in the image receptor for each of the projections. This is because there is no anti scatter grid built in the system[54]. For an improved cancer detection, optimization of these variables has been a subject of research into other techniques for breast imaging.

2.3 Breast CT

This modality was first proposed by Boorne et al in 2001[5][22] where they formulated the first image for clinical assessment[5]. Breast CT is an experimental imaging modality that employs projection data acquired with good angular sampling ranging from 0.25-1 degree [22]. With the limitations of mammography, such as the superposition of tissues and low sensitivity in dense breast, breast CT is developed to further reduce these limitations. The technique considerably improves the mass detection due to the flat-panel detectors and the cone beam system it employs. Studies have estimated that about 29-48 percent of concealed carcinoma are made visible in breast CT [58]. The modality can be useful for excellent scrutiny (staging) of established breast cancers, where a staging report from high resolution BrCT minimizes the number of positive limits (margins) in Breast Conservation Surgery (BCS). Unlike mammography where low energy X-ray are used, in contrast with BrCT 49-80kVp X-rays produced by tungsten target are consistently used[5][53]. The uncompressed breast is X-rayed and hundreds of projections are achieved with a field of view (FOV) wide enough to image the breast. The projection image is however poorly contrasted but the reconstruction process corrects the image[53][3]. Fig 2.4 is a scan of breast cancer lesion from a CT after reconstruction.
2.4 Photon Counting Breast Computed Tomography (PCBCT)

Currently, spectral photon counting breast computed tomography (PCBCT) has been under investigation by researchers with promising results in the use of the technology as a clinical device[56]. Mammography based on photoncounting detectors (PCDs) has proven to overcome the limitations of FFDM and DBT [52]. The device is accompanied by a scanner that uses standard polychromatic X-ray sources and dedicated PCDs that encompasses multi energy imaging. The PCDs are able to discriminate photons of different energy usually over a pre-defined set of energy windows [34]. Prototype based on cadmium Telluride (CdTe), cadmium zinc Telluride (CZT) and silicon semiconductor detectors are now commercially available, which allows accurate measurement of photon energies for clinical use[56]. Figure 2.7 is a schematic of a two-sliced spectral CT system.

Figure 2.4: CT scan showing breast cancer lesions [43]
with photon-counting CZT detector.

Figure 2.5: Schematic diagram of a two-sliced spectral CT system with photon counting CZT detector. Image adapted from [11]

Recently Ann-christin et al carried out a study that compared a PCBCT system with FFDM and DBT using surgical specimens, the confirmation of their results is tabulate in table 2.1

<table>
<thead>
<tr>
<th>systems</th>
<th>Sensitivity for microcalc (percent)</th>
<th>Specificity for microcalc (percent)</th>
<th>Sensitivity for lesion (percent)</th>
<th>Specificity for Lesion (percent)</th>
</tr>
</thead>
<tbody>
<tr>
<td>FFDM</td>
<td>82</td>
<td>71</td>
<td>45</td>
<td>76</td>
</tr>
<tr>
<td>DBT</td>
<td>70</td>
<td>75</td>
<td>62</td>
<td>62</td>
</tr>
<tr>
<td>PCBCT</td>
<td>85</td>
<td>83</td>
<td>65</td>
<td>76</td>
</tr>
</tbody>
</table>

Table 2.1: Comparism of FFDM, DBT and PCBCT [52]

The benefits of photon counting have previously been investigated, its use as a clinical devise has been hindered by the requirement of X-ray imaging, considering that quite a good number of photons are meant to be detected in a short time of image acquisition. Also that large number of data are meant to be transferred to the computer and kept for use. However, the advent of chips in electronics have made it feasible for an efficient process. The ability to undertake spectral decomposition of data is a major potential
advantage of photon counting CT compare to conventional CT. This allow better discrimination of different tissue types and hence might be used to enhance the contrast of lesions within the image. These processes are carried out with a greater consideration of how much dose is being absorbed by the breast tissues.

2.5 Importance of Dose in Computed Tomography

The persistent proposals of computed tomography CT as an imaging technology has been of concern recently[11], this is due to the fact that these technologies induce cancer during mammography screening, and have been a risk to women who undergo mammography screening, Hence its clinical investigation. Radiation dose received during mammography is therefore a key issue as a result of the risk of induced cancer due to population exposure. Although adipose tissues are not at a higher risk as glandular tissues, care is taken to monitor the amount of radiation dose received by the glandular part of the breast because of its high radio sensitivity[41]. Improvements in sensitivity according to Boone et al 2001 can be achieved with clinical CT as it completely removes the overlap of anatomical structures that have immense effect on mammography.

Despite this advantage, there exist differences in X-ray attenuation between tissues of breast particularly glandular and tumour tissues[41]. Dose assessment studies are required in breast imaging in order to determine accurately the amount of radiation dose involved during clinical application of CT [33]. It has been demonstrated [33] that the use of adhoc reconstruction CT algorithms could have effect on deposited dose. Other works of[40] also illustrates a quick and accurate procedure for estimating dose using synchrotron based studies on breast samples. Their work was aimed at establishing dose delivered in breast CT with the aid of Monte Carlo (MC) simulations and to also establish photon energy for dose minimization in CT which was based on monochromatic X-rays. Results suggests that CT imaging using monochromatic synchrotron radiation beams will allow assessment of an average dose with good accuracy in pre-clinical breast imaging.

Based on the risk associated with radiation especially during mammography screening as the breast is exposed to ionizing radiation, dose determination during CT exami-
nination is important as it measures risks to patients and attempts to reduce radiation without compromising the quality of diagnostic information. Methods such as reducing the peak tube voltage, using automatic exposure control and shielding can be useful, these methods could lead to sub optimal results.

### 2.5.1 Dose, Signal to Noise Ratio and Spectral Resolution in CT

Clinically, the quality of image in CT scan has become an issue as ways to reduce radiation dose to patients has become a priority in diagnostic imaging[2]. Obtaining images using a CT scan practically depends on the attenuation coefficients $\mu$ of different tissues. For every pixel constituting the end result of an image, a gray scale corresponds to the $\mu$ and linked to the Hounsfield scale. The detector turns around the breast to enable the determination of different $\mu$, signals are obtained from the detector at given angles which corresponds to projections. The projections in turn corresponds to sinograms of CT image. These sinograms are then reconstructed to obtain images that are targeted to be real images[2].

Unwanted change in pixel values of an image can be seen as noise in CT, technically referred to as quantum mottle. Noise in CT is measured through signal to noise ratio SNR where photons are compared to deviations of pixels. In general, the number of photons absorbed during CT imaging results to a greater SNR and lower noise in the image [2]. Application of intensifying screens with high kV reduces the amount of photons and radiation dose to patients and minimizes SNR. Radiation dose to patients is increased when mAs increases, and also determines SNR in CT. Slice thickness and patient size are also contributing factors, when dose increases in CT scan, the amount of noise decreases and subsequently contrast resolution is improved. Slice thickness on the other hand has a direct relationship with the number of photons that generates an image during scan. If slice thickness is reduced, the volume of tissue is reduced known as voxels. Smaller voxels absorbs less total number of photons, thereby increasing SNR. This is because of statistical nature of photon interaction. However an increase in slice thickness decreases spatial resolution. This also means that the size of tissue determines the amount of photons absorbed. If more photons are absorbed, less photons get to
Chapter 2. Review of Imaging Technology for Breast Cancer Screening

the detector, hence a reduction in SNR.

2.6 Principles of Mammography

This section examines some of the physical factors guiding mammography imaging system. In section 1.1.1 and 1.1.2 Breast Anatomy and Breast cancer were discussed as an introduction to the principles.

2.6.1 Mammography X-rays imaging system

Production of X-rays takes place in the X-ray tube containing a source (cathode) and a target electrode (anode). With the application of high voltage, electrons gain kinetic energy known as thermal excitation as they move from cathode to anode. The electrons hit the anode and energy is release in form of X-rays. This may be bremsstrahlung (continuous X-rays) and characteristic radiation, yielding characteristic peaks superimposed on to the continuous spectrum. The electromagnetic waves, typically the X-rays consist of photons with frequency $f$ and wavelength $\lambda$.

$$E = hf = \frac{hc}{\lambda}$$

where $h$ is planks constant, and $c$ is the speed of light. The wavelength has the order of Angstroms ($10^{-10}$) and the corresponding photon energies with the order keV

2.6.2 X-ray spectrum

A low energy x-ray spectrum is required in mammography to optimize contrast as the breast is unique in offering soft tissue without any bone anatomy presence. This is achieved by using molybdenum or rhodium target and a K-edge filter of molybdenum [11]. The X-rays tube may have more than one target and a filter wheel, which can vary the X-ray spectrum automatically. This is shown in figure 2.6
2.6.3 Interaction of X-rays Beam with tissue

X-rays are ionizing waves which ionizes an atom by releasing electrons. X-rays from X-ray tube irradiate the breast and photon energy is attenuated as it passes through breast tissue[41]. Each interaction process reduces the X-rays photons from the beam of X-rays by absorption or scattering away from the breast tissue. In principle, interaction of X-ray photon is based on photoelectric absorption, Raleigh scattering and Compton effect. During photoelectric absorption, photon interacts with the breast tissue and it is completely absorbed, the energy of photon is then transferred to electron and subsequently ejected from the atom. Most photon absorption in the breast tissue is as a result photoelectric absorption which influences attenuation[41].

Raleigh scattering also known to be coherent, where X-rays are scattered by atomic electrons. This process neither excited nor ionizes an atom because energy of scattered photon is same with the incident X-ray.

In Compton effect the interaction process occurs in between incident ray photon and a loosely bound electron of an atom of the breast which is the absorbing material. Part
of the X-rays energy is transferred to the shell of the atom of breast tissue by deviating through an angle. Ionization is caused when the electron is ejected.

These processes described above reduces the intensity during interaction with a material of thickness $d$ in various ways which are statistical in nature and are characterized by a fixed probability. The sum of these probabilities is the probability per unit path length, an X-ray photon is removed from the beam. The strength of the exiting beam obeys the exponential law:

$$I_{out} = I_{in}e^{-\mu d}$$  \hspace{1cm} (2.2)

where $\mu$ is the attenuation coefficient with a unit of $cm^{-1}$ and is a function of both the photon energy and the material. $I_{in}$ is the incoming beam intensity and $I_{out}$ is the exiting beam. Equation 2.2 is considered during acquisition of image in a compressed breast.

### 2.6.4 Breast compression

In the process of image acquisition, the compression paddle as used in mammography reduces motion blur, dose scatter and improves contrast. It also enables a softer x-ray spectrum and reduces unsharpness, dynamic range requirement with a reduction in the amount of tissue overlap which allows good visualization of breast tissues[11].
2.6.5 Anti Scatter Grid

Scattered radiation during image acquisition reduces the contrast of an image, this depends on the photon energy, breast size and the image receptor that records the scattered radiation. This can be measured using contrast degradation factor (CDF). CDF decreases with an increasing breast thickness and it is the ratio of image contrast with and without the effects of scatter [11]. The grid is made of parallel lead strip and placed between breast support and image receptor as in figure 2.6. With the passage of primary photons through the grid, the scattered photons are absorbed, usually 75-80 percent in mammography grids [6].

2.6.6 X-ray Receptors

Images of X-rays are seen with the aid of image receptors which is a technology that transforms X-ray beam into a visible image. X-ray receptors have to be larger than the part to be imaged for proper visibility of the image and should have the ability to detect target objects with emphasis that the size and contrast of object is limited to quantum statistics [47]. Processes for image reception differs for screen film system and digital system. Unlike screen film system, in digital systems, x-rays are absorbed and multiple secondary electrons in a photo conductor are released in digital systems and subsequent drifting of electrons and holes takes place [47].

In general, X-ray photons must interact with receptor materials as seen in figure 2.7 for signal production. Quantum Efficiency for an X-ray of Energy $E$ is given by equation 2.3.

$$A_Q = 1 - \exp \left\{ -\mu (E, Z) T \right\}$$

(2.3)

where $\mu$ is linear attenuation coefficient of receptor material, $Z$ atomic number of material and $T$ is thickness of material. $A_Q$ is highest at low energies as it decreases with increasing energy, it is also known to be a function of Energy or an effective value over a spectrum of X-rays. Images generated by X-ray photons are statistical in nature, this means that images can change about the mean predictive value (MPV) [32].
Figure 2.7: Image receptor located behind a detector where the X-ray exposure is measured just before it enters the image receptor for automatic exposure control system[47].

In as much as these principles have worked towards the progress of breast imaging, photon counting spectral CT has been under investigation recently and has been proposed for breast mammography screening especially towards calcification and lesion visibility[10].

2.7 Calcification and Lesions

It is not certain or clear about the various desposition of calcium within the breast. However, studies have shown that normal and abnormal body functions results to breast calcification[10]. Breast calcification is found in 85-95 percent of patients who suffers from ductal carcinoma insitu (DCIS)[10]. The appearance of microcalcification as bright spots on mammograms as seen in figure 2.8 is as a result of high X-ray
absorption of calcium. Location, distribution size, density and number are used in assessing microcalcification and these are used as determining factors on which to base a decision of malignancy or benign disease[51].

Figure 2.8: Fine microcalcifications in the upper outer quadrant of the right breast. Image adapted from[7]

Mammography and other techniques can conveniently be used in diagnosing benign breast disease which constitutes different kinds of lesions such as Mastitis acute Mastitis and Granulomatous Mastitis, hence a various range of symptoms, some of which includes inflammation, nipple retraction or inversion, nipple discharge or skin changes[27].
2.8 Contrast in Mammography

Clinical studies have demonstrated that screening mammography reduces breast cancer mortality [15]. The effectiveness of mammography can only be seen by the ability to identify tumours in breast cancer depending on the absorption of X-rays in the affected tissue, different from adipose and glandular tissues. Cancerous tissues in breast appears bright with similar intensity to granular tissues which makes detection difficult during imaging, particularly in the dense breast [30],[5]. At present screening mammography aims to reduce false positive results and to resolve the limitations of superimposed tissues in some technologies. This can be achieved if the magnitude of the signal difference between the tumour and its surrounding background is considered, which is the contrast.

In breast imaging, high contrast is sought to enable differentiation of normal structures from pathological structure. The contrast also enables the detection of calcium deposits in the breast. A cancerous breast contains the lesion and other surrounding tissues as a result, the contrast is caused by the differences in X-ray attenuation properties of the lesion and the surrounding tissues. The thickness of the lesion is also a depending factor[59].

2.9 Contrast and Radiation Dose

Patients who routinely undergo mammography screening need to be protected against accumulated radiation dose from mammography screening. When ionizing radiation interacts with the tissue, some level of radiation dose is absorbed. Although radiation dose from mammography equipment are low, usually 0.4$mSv$ for adult approximate effective radiation dose[11]. Clinical studies shows that mean granular dose (MGD) from dedicated breast CT compares with FFDM [48]. Risks of radiation dose depends on the long term exposure to radiation involving routine examination. For a certain lesion detection level, minimal dose is used in order to minimize the dose to the entire screening population. Whilst risk to the individual is very low, the key statistics needed are the number of cancers induced per 1000 women screened as a result of a
single screening session across the entire screening population and the accumulated risk associated. Therefore the primary risks in mammography is the resultant across a screening population as a result of irradiation of fibrogranular tissue to ionizing radiation[28].

During mammography screening emphasis is made on choosing an x-ray beam spectrum that provides a balance between contrast and dose which depends on photon energy as shown in figure 2.9. At low energies contrast is high but dose at this level is low, this is because penetration is low at these levels. As the body section penetration increases at high photon energies, dose is reduced but the contrast becomes increasingly poor because of the deleterious effect of Compton scattering which increases with energy.

2.10 Variation of Dose in Computed Tomography

Accurate radiation doses to patients during clinical examination using CT are necessary and the variation of dose with X-ray energy is useful [60]. This is to measure radiation dose to patients and as well compare to the X-ray energy as it decreases or increases. Radiation dose absorption in projection radiography and mammography differs from absorption in CT in many cases as dose in mammography decreases with increasing energy where as patient doses increases with energy in CT. In mammography, the Automatic exposure control sustains the exposure for receptors to receive adequate dose to register an image at a low noise level. However, the speed of rotation in CT which is basically the exposure time is fixed, as a result X-ray tubes are more efficient at higher energies which also implies more emitted photons. These photons increases the efficiency of X-ray production. Therefore patient dose increases as energy increases for fixed exposure time.
Figure 2.9: General Relationship of Contrast and Dose to Photon Energy in Mammography [62].
Chapter 3

Methodology and Simulation Framework

This chapter describes simulation work in Breast CT and discusses the method through which the data presented in chapter 4 has been realized using in-house simulation software. The aim of this experiment is to investigate whether a spectral imaging approach (for example, using photon counting detectors) in breast CT can render lesions more visible than using a conventional CT approach that relies on integrated signals.

3.1 Previous work on Lesion Simulations

Simulation work in the studies of medical imaging deals with the image quality with phantoms and test objects that agrees with the requirements of each study[4]. Computer simulation provides affordable and flexible process for undergoing image quality assessment and X-ray simulation is specific to particular radiographic technology and the geometry of acquisition.

Early works in breast lesion simulation were dependent on Gaussian profile[8][45]. Although these methods could provide adequate mathematical analysis, it could not represent realistically the lesions due to its complexity[8]. Lesion simulation involves the use of mathematical models as masses inserted into clinical images. Some authors employ the use of spheres and ellipsoids to simulate masses[4].
Recently, Rashidnasab[51] published a new method of generating 3D masses using Diffusion Limited Aggregation (DLA), this masses were inserted into 2D mammograms using a physics based approach as shown in figure 3.1.

![Simulated DLA masses in 3D and corresponding projections in 2D image](image.png)

**Figure 3.1:** simulated DLA masses (a) to (c) in 3D, (d) to (f) and the corresponding projections in 2D image adopted from[51]

The generation of these simulated DLA masses according to [14] was based on assigning a voxel of 3D binary matrix as a mass. Furthermore, a set of random particles were inserted from a given point known as launch site which was independently processed. The process of random walk (RW) was considered where particles are expected to cluster with a probability called sticking probability. Particles are however not useful to the process once they fail to satisfy this probability. If however the probability is met, clusters are formed by additive process of the random particles. Simulation has therefore been used to achieve desired results in breast imaging system as mentioned in the studies cited above.

### 3.2 Breast phantom Design

The phantom used in this work has an approximate diameter of 100mm with two lesions inserted each measuring 5mm, representing 100 percent glandular tissue, and a calcification clutter with an average calcification diameter of 0.1mm. The breast
phantom explicitly model tissues including; adipose, glandular, blood vessel, cooper ligament and air. The attenuation of each tissue changes for each energy as the energy changes along the polychromatic beam. The two dimensional computational phantom shown in figure 3.2 was built for the development, optimization and evaluation of spectral imaging system aimed at reproducing breast characteristics for this work. The idea was developed to replace the physical phantom which are until now expensive. Computer Aided Software (CAS) developed by Elangovan [14] and Rashidnasab[51] was used, where the insertion cite for calcification clutter and lesion inserted manually. To further experiment on the developed phantom to determine the contrast at different photon energies using arbitrary energy bins, a code was developed for the simulation work see Appendix E. This code generates sinograms during simulation and images which are reconstructed to enable the determination of contrast at different photon energies. Similarly, a code has been generated to calculate the absorbed dose at these energies as illustrated in appendix E.

The method of Elangovan etal[14] was also useful in the development of this phantom, they worked on the method of generating a quasi-realistic voxel phantom. The phantom was used in simulating DBT and mammography. The simulation process was carried out by generating an empty breast volume designated by different tissue structures. Furthermore, granular tissue, cooper ligament and blood vessels were simulated. These simulated radiological images shown in figure 3.3 were validated against real images. The process of developing the phantom in figure 3.2 was undertaken based on optical appearance and location in comparison with areas that malignancies could be found.

3.3 Siddon Algorithm

The Siddon algorithm[57] is a 3-D ray tracing algorithm aimed at calculating the exact radiological path passing through voxels. Unlike ray tracing used in computer vision, which is used for calculating surface effects, the siddon algorithm is used for X-ray tracing through the volume of objects and calculating the attenuation, or loss of photon intensities through a mixture of different materials/intensities. The total attenuation of the tissues is measured along the X-ray projection passing through the tissues as
Figure 3.2: Breast Phantom.

shown in figure 3.4. Each voxel $v_1$, $v_2$, $v_3$ and $v_4$ has different values of attenuation shown in equation

$$I = I_0 e^{-(\mu_1 + \mu_2 + \mu_3 + \mu_4 + ... + \mu_N) d}$$

(3.1)

$$\sum_{i=1}^{N} \mu_i = \frac{1}{d \ln I_0}$$

(3.2)

Where: $d$ is the width of voxel, $\mu$ is the attenuation coefficient, $I$ is the X-ray flux exiting the phantom and $I_0$ is the X-ray flux entering the phantom. The distance enclosed by each voxel is the difference between two adjacent parametric values. The product of the distance in each voxel and the densities are summed up to give the radiological path [6].
3.4 Filtered Back Projection

Because of its simplicity and computational efficiency, filtered back projection which is a standard reconstruction method is still used clinically. The working process is in two steps namely, filtering of data and back projection of the filtered data. In image acquisition which is 2-dimensional, each row of projection plays a distinct role by representing a sum total of counts on a straight line through the subject to be imaged. The process of back projection comes in by repeating the total sum of counts at every point along the same line in opposite direction with respect to its origin. For all pixels and angles involved the process is repeated. These process however follows with some
Figure 3.4: A representation of different tissues with different attenuation in a phantom artifacts and image blurring which is attributed to the limited number of projections. Avoiding this limitation means filtering the projection before back projecting along the same lines[37].

The ramp filter plays an important role of limiting the blurring of images by stopping the low frequencies that could have otherwise appear in the image and is known to be a high pass filter. This process eliminates the artifacts caused by the simple back projection process and in return sharpens areas of image that changes the signal as a result of the blurring which is essentially the edges of the images[26]. However, high pass filter increase statistical noise but the work presented in this thesis is independent of photon/statistical noise. Figure 3.5 is a simple representation of filtered back projections while figure 3.6 demonstrates how projections are acquired by performing a Radon transform demonstrated in equation 3.3.
3.4. Filtered Back Projection

Figure 3.5: An illustration of filtered back projection (a) Acquisition of three projections (b) Back projected projections (c) Filtered back projected Projections [37]

Figure 3.6: parallel beam geometry

\[ P(\rho, \theta) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} I(x, y) \delta(x\cos\theta + y\sin\theta - \rho) \, dx \, dy \]  \hspace{1cm} (3.3)

Where; \( \delta(x\cos\theta + y\sin\theta - \rho) \) is the line through the object, \( P(\rho, \theta) \) is the projection image, \( \rho \) is the detector at orientation \( \theta \) and \( I(x, y) \) is the density function of an object[66].

Further more, a geometrical model for fan beam data acquisition is presented in figure 3.7 where fan beam projections are first weighted by \( \cos\theta \) (cosine of the fan beam angle). Secondly weighted data are convolved with a ramp filter \( h \) with respect to the fan angle \( t \), which is aimed at filtering out the low frequencies and passing the high frequencies as earlier discussed in this section. Back projecting the filtered projections over 360 degrees
using the inverse square distance from the focal point to voxel of interest results in the
reconstructed images and scaled by \( R/2 \), where \( R \) is the distance from the sample to the
source [66]. The work of this study took advantage of the relationship between Radon
and Fourier transform for accuracy of results. The effects of blurring was avoided and
distance between source of X-rays and phantom was analyzed using direct fan-beam
reconstruction algorithm through FBP for differential phase contrast.

![Geometrical model for fan-beam data acquisition](image)

**Figure 3.7:** Geometrical model for fan-beam data acquisition

### 3.5 Simulation Framework

The process for which simulation work for this study is carried out has been described
using key components of the simulation frame work presented in figure 3.8. A series of
in house simulation tools were used to simulate a photon counting breast CT system.
A polychromatic energy spectrum was also generated from the in house software based
on the energy spectrum considered for these experiments. The geometry considered
was 840mm and 740mm which are the distance from source to detector (SDD) and
source to sample (SOD) respectively. In addition, the anode filtration combination
used was Rh/Al. This simulation was taken to represent a CT during which a ray
tracing algorithm was used to trace the photons. A phantom as previously described
in section 3.2 was then used along with the siddon algorithm and Fan beam acquisition
process to generate sinograms. Reconstructed Images were processed from the generated
3.5 Simulation Framework

sinograms. In the reconstruction process, an angular beam spacing of 0.5 degree, fan sensor spacing of 1 cm and a line fan sensor geometry were considered.

![Block diagram of the key components used within the simulation framework.](image)

Figure 3.8: Block diagram of the key components used within the simulation framework.

### 3.5.1 Photon Energy Investigation

In order to examine whether photon counting technology can be used to enhance lesion contrast, photon energies between 0 and 120keV were investigated initially using bins of 20keV each in a first experiment, then using bins of 10keV in a second experiment and finally in bins of 5keV in a third experiment. The largest bin width as an initial experiment and the final test was used to look at any fine structure or other effects that may have been revealed. A region of interest (ROI) measuring 4x4 pixels was placed in the lesion and referred to as 

\[
ROI_l
\]

Similarly 4x4 ROIs were as used for the background in four different positions and the average value estimated, referred to 

\[
ROI_{bg}
\]

The contrast was then calculated using equation 3.4 and 3.5 for lesion and calcification respectively

\[
Contrast = \frac{(ROI_{bg}) - (ROI_l)}{(ROI_{bg})} \tag{3.4}
\]

\[
Peakcontrast = \frac{(ROI_{bg}) - (Max_{calc})}{(ROI_{bg})} \tag{3.5}
\]

where; \(Max_{calc}\) is the maximum calcification peak contrast
The ability to undertake spectral decomposition of the data is a major potential advantage of photon counting CT compared to conventional CT. This may allow better discrimination of different tissue types and hence might be used to enhance the contrast of lesions within the image. In view of this, the lesion contrast at each energy bin was determined using equation 3.1. Similarly equation 3.2 was used in determining the Peak contrast for the calcification clutter due to the small sizes. This was further matched with the linear attenuation coefficients of glandular tissue, fat and calcification. This is to identify the energy bin and the corresponding attenuation between the tissues.

3.6 Image Reconstruction

The whole essence of image reconstruction is to appraise a spatial dissemination of some parameters in an object from its projections. These projections are a set of measured boundary intrinsic (integral) values of the parameters. The parameters here is attributed to the linear attenuation coefficient for X-ray transmission[5]. For the purpose of this work the Filter back projections (FBP) method is considered. To improve the image quality of FBP ramp filters are used but regulated with some parameters that must be suitable to the given acquisition. To a greater extent, in FBP algorithm ramp filters are modelled based on the frequency sampling denseness.
Chapter 4

Results and Discussion

The previous chapter described Breast CT simulation methodology used in this thesis and the method used in carrying out experiments. Results of experimental findings of analysing contrast across a range of energies are presented in this chapter with selected reconstructed images for these experiments. Furthermore, outcome of lesion and calcification decomposition for each experiment at different energy bins and the standard approach to CT are tabulated and analyzed graphically. Results of findings from the experiments are also discussed here with emphasis on the lesion and calcification decomposition pointing out reasons and observations for the outcome. The results of experimental procedure follows with data generated from simulation work. Figure 4.1 describes the phantom used in this work which is built based on the method described in Elangovan et al[14], previously summarized in section 3.2. The main tissue classes simulated as seen in the phantom above are; glandular tissue, calcification clutter, adipose, blood vessels, and Cooper’s ligaments. As earlier mentioned colours have been inserted to clearly distinguish tissues inserted in the phantom. The attenuation coefficients assigned to these tissue classes varies at each energy point as it changes along the energy spectrum. Figure 4.2 is the plot of attenuation coefficients of the main tissues concerned with, namely, calcification, fat and glandular tissues. The Siddon Algorithm described in section 3.3 was then used to simulate the resulting sinogram for 840mm distance from source to detector and 740mm source to sample using the imaging geometry shown in figure 3.7. This represents an expected results that may
be observed in X-ray CT scanner using charged integrating detectors. However the work featured in this thesis is concerned with the potential development of breast CT using photon counting technology. The investigation therefore begins with considering the appearance of the sinogram and resulting reconstructed images at various discrete energy windows. Figure 4.4 to 4.6 shows various contrasts seen at these different energy bins with figures 4.4a and 4.4b representing reconstructed images at bin width of 20keV. Figure 4.5a and 4.5b represents reconstruction at bin width of 10keV while 4.6a and 4.6b illustrate images at 5keV.
Figure 4.2: Attenuation plots of calcification, fat and glandular tissues.
Figure 4.3: X-ray spectrum across photon Energies.
Figure 4.4: Reconstruction with bin width of 20keV at [a] (1-20)keV and [b] (101-120)keV
Figure 4.5: Reconstruction with bin width of 10keV at [a] (1-10)keV and [b] (111-120)keV
4.0.1 Lesion Decomposition with Integrating Detector Technology Approach

Investigations carried out on lesion contrast using the integrating detector approach, was based on the in house software spectrum which represents the spectrum in a conventional CT system. The phantom as described in figure 4.1 was simulated using siddon algorithm over a polychromatic range of 1-120keV with the imaging geometry described in 4.0.1. Figure 4.7a is the result of the reconstructed image with the corresponding sinogram shown in figure 4.7b. Further analysis of the reconstructed image was then based on the lesion and calcification decomposition where contrast for both lesion and calcification were investigated for both the sinogram and the reconstructed image with respect to their backgrounds. This was to observe if there was any difference in contrast. Unsurprisingly, there was no significant difference in the percentage contrast for both sinogram and reconstructed image as observed in the decomposition of the lesion and calcification. Results obtained from the work of this reconstruction shows percentage contrast of 6.83 percent for lesion and 64.5 percent for calcification.
Because the ability to undertake spectral decomposition is a major potential advantage of the photon counting approach compared to conventional CT, further investigations were carried out using the photon counting approach where photons conveying a range of energies are aggregated into a set of discrete energy windows or bins. This process is to acquire data in different energy windows and identify individual windows with the corresponding contrast and spectral energy values.
4.0.2 Investigating Lesion Decomposition at Bins of 20keV

Seven arbitrary energy bins of 20keV each in the first experiment were selected, this was to investigate the decomposition at each energy bin and investigate contrasts at each bin. To analyze the contrast, measurements taken were based on the method described in section 3.5.1 and equation 3.1 employed. Results presented in figure 4.8 shows the corresponding exponential drop in contrast with the energy bins where contrast decreases with increasing photon energy. This action is attributed to the difference in linear attenuation coefficient of adipose and glandular tissues. Results of this experiment as graphically presented in figure 4.8 demonstrates change in contrasts and that decomposition of lesion in bins provided appreciable contrast following a pseudo-exponential relationship between contrast and energy bins, with 22 percent in bin (1-20)keV and 5 percent in bin (61-80)keV compared to the polychromatic decomposition where the lesion contrast was 7 percent as presented in table 4.1. This percentage is closer to 8 percent contrast of the lesion decomposition corresponding to energy bin (61-80)keV. However, decomposition for photon counting energies for bin (81-100)keV and (101-120)keV, respectively 2.5 percent and 1.6 percent were comparably lower than polychromatic decomposition of 6 percent. This indicates that polychromatic contrasts could be higher than the contrasts in the higher energy region, but that there is a distinct advantage to focusing imaging on the medium to lower energies.
Figure 4.7: (a) Reconstruction from polychromatic projections. (b) Sinogram from the phantom using polychromatic projections.
Figure 4.8: graphical representation of lesion contrast versus Energy at bins of 20keV each
Table 4.1: Lesion contrast corresponding to Energy bins at bin width of 20keV with 1-120 representing polychromatic beam

<table>
<thead>
<tr>
<th>Energy Bin(keV)</th>
<th>Lesion Contrast (Percentage)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-20</td>
<td>22</td>
</tr>
<tr>
<td>21-40</td>
<td>15.6</td>
</tr>
<tr>
<td>41-60</td>
<td>7.8</td>
</tr>
<tr>
<td>61-80</td>
<td>4.4</td>
</tr>
<tr>
<td>81-100</td>
<td>2.5</td>
</tr>
<tr>
<td>101-120</td>
<td>1.6</td>
</tr>
<tr>
<td>1-120</td>
<td>6.83</td>
</tr>
</tbody>
</table>

4.0.3 Investigating Lesion Decomposition at Bins of 10keV

In order to carry out further investigations on the photon counting approach, another twelve arbitrary energy bins of 10 keV each has been investigated, this was to examine if different choices of energy bins might improve the lesion contrast. Results obtained and analyzed in figure 4.9 demonstrates the change in contrast and indicates a similar exponential drop in lesion contrast as recorded in the 20keV bin width experiment, with 21 percent contrast at the lowest energy bin of (1-10)keV and 1.2 percent at the highest energy bin of (111-120)keV as shown in table 4.2. Differences in the percentage values compared to the first experiment is attributed to the energy bin width and the rate at which different tissues were attenuated. Similarly, the polychromatic contrast of 7 percent was again less than the first four energy bins with 8 percent being the closest contrast and corresponding to (41-50)keV. The polychromatic contrast is appreciably higher than the contrast in the energy bins from 51 down to 120keV.
Figure 4.9: graphical representation of lesion contrast versus Energy at bins of 10keV each
Energy Bin(keV) | Lesion Contrast (Percentage)
---|---
1-10 | 21
11-20 | 16.5
21-30 | 14.5
- | -
- | -
91-100 | 1.6
101-110 | 1.4
111-120 | 1.2
1-120 | 6.83

Table 4.2: Lesion contrast corresponding to Energy bins at bin width of 10keV with 1-120 representing polychromatic beam

**4.0.4 Investigating Lesion Decomposition at Bins of 5keV**

Further experiments were carried out by reducing the the energy width to 5keV each thereby having twenty four energy bins in total, this was to re-examine lesion decomposition in much more smaller bin widths. Investigation of lesion decomposition in each of the bin shows an exponential relationship between contrast and energy bin with 20 percent being the highest contrast at bin (1-5)keV and 0.9 percent at bin (116-120)keV as shown in figure 4.10. In the same trend lesion decomposition as investigated using the standard approach to CT was less than the first four energy bins with the closest lesion contrast at bin (31-35)keV. However, polychromatic contrast was higher than the contrasts from 31keV down to 120keV see table 4.3. The trend as observed in all three experiments corresponds to attenuation of glandular and calcification tissues as shown in figure 4.1. Having investigated lesion decomposition at arbitrarily different energy bins and compared to the polychromatic investigation for CT standard approach, calcification decomposition at same energy widths were also investigated. This was to understudy the percentage peak contrast of calcification compared to their background as seen in figure 4.1
Figure 4.10: graphical representation of lesion contrast versus Energy at bins of 5keV each
### Table 4.3: Lesion contrast corresponding to Energy bins at bin width of 5keV with 1-120 representing polychromatic beam

<table>
<thead>
<tr>
<th>Energy Bin(keV)</th>
<th>Lesion Contrast (Percentage)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-5</td>
<td>20</td>
</tr>
<tr>
<td>6-10</td>
<td>18.3</td>
</tr>
<tr>
<td>11-15</td>
<td>4</td>
</tr>
<tr>
<td>16 –</td>
<td>–</td>
</tr>
<tr>
<td>–</td>
<td>–</td>
</tr>
<tr>
<td>106-110</td>
<td>1.2</td>
</tr>
<tr>
<td>111-115</td>
<td>1.2</td>
</tr>
<tr>
<td>116-120</td>
<td>0.9</td>
</tr>
<tr>
<td>1-120</td>
<td>6.83</td>
</tr>
</tbody>
</table>

In the current study, emphasis is also made on the radiation dose to patients as it is a very critical component in CT when used clinically as a modality for breast screening. This is because of the high radio-sensitivity of glandular tissue. A code generated for simulating dose has been provided and the variation of dose across energy bins have been studied. All simulation have been performed using an in house software which covers photon energy range of 1-120keV. The aim of this experiment was to investigate the radiation dose delivered as the photon energy increases, and also estimate photon energy monochromatic X-ray where relative dose can be considered. As described in this thesis, breast phantom used consist of lesions, calcification, glandular tissue and fat with different attenuation coefficients. Dose estimation here depends on the number of photons absorbed and considered in this thesis as relative dose. This is because dose can not be purely based on the number of photons absorbed, but calculated by the number of photons multiplied by the absorbed photons. This is an idealized calculation which is based on parallel beam geometry where photons removed from beam are assumed to be absorbed locally rather than lost in the surrounding air. Results presented in figure 4.11 suggests that the best energies for clinical CT examination is between 50-65keV.
Based on the investigation carried out in this work, contrast and dose as seen as a function of photon energy where contrast reduces as photon energy increases. Also exponential attenuation of X-ray beam results in lower dose while lower photon energy results in higher contrast and subsequently higher doses.
Figure 4.11: graphical representation of relative dose and contrast versus Photon Energy
4.0.6 Calcification Decomposition with Integrating Detector Technology Approach

The phantom shown in figure 4.1 was used to analyze calcification peak contrast based on the conventional CT approach using the in house software with a polychromatic energy of 1-120keV. This investigation was facilitated with the imaging geometry described in section 4.0.1 and simulated using the Siddon algorithm as described in chapter 3. Figure 4.6 illustrates the image of the reconstruction. Results of this decomposition is presented in table 4.4 where the calcification peak contrast was 64.5 percent. In order to compare contrasts using the conventional CT approach and the spectral imaging approach based on the polychromatic energy, further analysis were performed using the photon counting approach where three sets of experiments were carried out and photons aggregated into bins of 20keV, 10keV, and 5keV as previously examined in sections 4.02-4.0.4. This is to investigate percentage peak contrast using the method described in section 3.5.1 and presented in the following sections.

4.0.7 Investigating Calcification Decomposition at Bins of 20keV

For uniformity in experimental process, the same energy width of 20keV was used in investigating calcification decomposition at each energy bin of seven energy windows. Results presented in figure 4.12 demonstrates an exponential decline in contrast with increasing energy bins. Results indicates that 95 percent peak contrast aligned with the lower energy bin of (1-20)keV and 39 percent at the higher energy bin of (101-120)keV. The 65 percent contrast obtain during during the investigation of lesion calcification for conventional CT was however lower than contrast for photon counting approach at lower energies from 1-100keV and higher from 101-120keV see table 4.3. This is attributed to the attenuation of the tissues as described by the attenuation plots in figure 4.2
Figure 4.12: graphical representation of calcification contrast versus Energy at bins of 20keV each
Table 4.4: Calcification peak contrast corresponding to Energy bins with 1-120 representing polychromatic beam

<table>
<thead>
<tr>
<th>Energy Bin(keV)</th>
<th>Calcification Peak Contrast (Percentage)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-20</td>
<td>95</td>
</tr>
<tr>
<td>21-40</td>
<td>93</td>
</tr>
<tr>
<td>41-60</td>
<td>83</td>
</tr>
<tr>
<td>61-80</td>
<td>84</td>
</tr>
<tr>
<td>81-100</td>
<td>78</td>
</tr>
<tr>
<td>101-120</td>
<td>39</td>
</tr>
<tr>
<td>1-120</td>
<td>64.5</td>
</tr>
</tbody>
</table>

4.0.8 Investigating Calcification Decomposition at Bins of 10keV

Investigating calcification decomposition and subsequent contrast evaluation was considered at a lower energy window of 10keV bin width in order to compare calcification peak contrast with the result of the experiment in section 4.0.6, whether there could be any significant change in calcification peak contrast with regards to energy bins. Results of experiment with 10keV bin width in fig 4.13 shows a decrease in contrast with increasing photon energy which corresponds with the trend of attenuation plot as illustrated in figure 4.2. Calcification peak contrast evaluated at this bin width shows significant improvement when compared to the 65 percent peak contrast investigation from the conventional CT approach due to the inability of the conventional CT approach to utilize adequately the property of spectral decomposition. Tabulated result of these investigations are presented in table 4.5. However investigation with the conventional CT had an appreciable calcification peak contrast of 65 percent more than the photon energies from bin 100keV down to 120 keV, this is attributed to the penetration of photons at this points
Figure 4.13: graphical representation of calcification contrast versus Energy at bins of 10keV each
<table>
<thead>
<tr>
<th>Energy Bin (keV)</th>
<th>Calcification Peak Contrast (Percentage)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-10</td>
<td>93</td>
</tr>
<tr>
<td>11-20</td>
<td>90</td>
</tr>
<tr>
<td>21-30</td>
<td>88</td>
</tr>
<tr>
<td></td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>–</td>
</tr>
<tr>
<td>91-100</td>
<td>73</td>
</tr>
<tr>
<td>101-110</td>
<td>42</td>
</tr>
<tr>
<td>111-120</td>
<td>36</td>
</tr>
<tr>
<td>1-120</td>
<td>64.5</td>
</tr>
</tbody>
</table>

Table 4.5: calcification peak contrast corresponding to Energy bins at bin width of 10keV with 1-120 representing polychromatic beam

### 4.0.9 Investigating Calcification Decomposition at Bins of 5keV

Re-examination of calcification decomposition at 5keV was considered to further analyze the possibility of having different peak contrast as was previously analyzed for lesion calcification in section 4.0.4. Graphical analyses of results presented in figure 4.12 demonstrates the exponential decline in contrast as photon energy increases along the spectrum. Shown also in figure 4.12 is the analyses based on the conventional CT where percentage contrast is 65 percent. Results also shows that analysis with photon counting demonstrate improved contrast visualization for photon energies bins between 1 to 70keV more that the 65 percent recorded in in the conventional CT approach as shown in table 4.6. However, percentage contrast for the CT approach was higher than the photoncounting contrasts from 71 keV. this trend is attributed to the inability of this approach to utilize all the data in the spectrum.
Figure 4.14: graphical representation of calcification contrast versus Energy at bins of 5keV each
<table>
<thead>
<tr>
<th>Energy Bin (keV)</th>
<th>Calcification Peak Contrast (Percentage)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-5</td>
<td>91</td>
</tr>
<tr>
<td>6-10</td>
<td>89</td>
</tr>
<tr>
<td>11-15</td>
<td>88</td>
</tr>
<tr>
<td>106-110</td>
<td>40</td>
</tr>
<tr>
<td>111-115</td>
<td>35</td>
</tr>
<tr>
<td>116-120</td>
<td>33</td>
</tr>
<tr>
<td>1-120</td>
<td>64.5</td>
</tr>
</tbody>
</table>

Table 4.6: Calcification Peak contrast corresponding to Energy bins at bin width of 5keV with 1-120 representing polychromatic beam.
Chapter 5

Conclusion and Further Work

5.1 Conclusion

The work of this thesis presents an image simulation framework that generated images from polychromatic energy spectrum characterized in this thesis as (1-120)keV and a photon counting approach which covered a range of arbitrary energies where photons combined into a set of discrete energy windows or bins, which has allowed the production of synthetic images. The work presented here represents an idealized case of noiseless images that do not contain scatter or photon noise in order to study the intrinsic properties of contrast in CT. As described in chapter three, a breast phantom of diameter 100mm was analyzed using a photon counting approach to simulate breast lesions generated by the method described by Rashidnasab et al [51] and calcifications. Investigations carried out in 6 different experiments for lesion decomposition and calcification decomposition has compared photon counting approach and standard conventional approach to CT. In all photon counting approach experiments for lesion decomposition described in chapter 4, higher contrasts were recorded between 1-60keV more than conventional CT approach while calcification decomposition recorded higher contrasts between 1-100keV. High contrasts values has been achieved at low energy bins (photon counting approach) which corresponds to the attenuation of glandular and calcification tissues. Photon counting approach has shown promise for the visualization of synthetic images based on contrast as described in this thesis. The result of the
experiments is attributed to the attenuation coefficients of tissues used in the phantom (described in chapter 3) with the varying energy along polychromatic beam.

In conclusion, contrast is highest at lower energies. This is where photoelectric absorption dominates and moreover where absorption in breast tissue is highest. However in order to form an image, the highest differential absorption is needed to generate contrast, but this competes with the need for maximum transmission to minimize dose to the patient, and also minimize the effect of dose limiting noise. Therefore, a medium range of spectral energies is ideal, between 40-60keV for optimizing contrast in spectral breast imaging.

5.2 Further Work

The work presented here represents an idealized case of noiseless images. More detailed Physics will be incorporated in the simulation work including scatter, which will apparently reduce contrast deferentially across different range of energies. Also, dose which will be seen as photon noise will be Incorporated as well. In addition the effect of breast motion will be considered in future work based on the data generated. The statistical effects of photon counting will also impact on the results presented here.
.1 Appendix A

The result of the experiment work carried out as described in chapter three for decomposition of lesion and calcification at a bin width of 20keV are presented herein.

<table>
<thead>
<tr>
<th>Energy Bin (keV)</th>
<th>Lesion contrast (percent)</th>
<th>Calcification Peak contrast (percent)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-20</td>
<td>35.4</td>
<td>95</td>
</tr>
<tr>
<td>21-40</td>
<td>22.1</td>
<td>93</td>
</tr>
<tr>
<td>41-60</td>
<td>18.4</td>
<td>90</td>
</tr>
<tr>
<td>61-80</td>
<td>7.7</td>
<td>84</td>
</tr>
<tr>
<td>81-100</td>
<td>2.5</td>
<td>78</td>
</tr>
<tr>
<td>101-120</td>
<td>1.6</td>
<td>39</td>
</tr>
<tr>
<td>1-120</td>
<td>6.83</td>
<td>64.5</td>
</tr>
</tbody>
</table>

Table 1: Lesion and calcification decomposition at bin width of 20keV
.2 Appendix B

The analysis of experimental results where lesion and calcification were decomposed at bin width of 10keV are hereby presented.

![Spectral Decomposition Table]

<table>
<thead>
<tr>
<th>Energy Bin (keV)</th>
<th>Lesion contrast (percent)</th>
<th>Calcification Peak contrast (percent)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-10</td>
<td>32.1</td>
<td>93</td>
</tr>
<tr>
<td>11-20</td>
<td>15.9</td>
<td>90</td>
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<tr>
<td>21-30</td>
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<td>88</td>
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<tr>
<td>31-40</td>
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<tr>
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<td>7.8</td>
<td>86</td>
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<tr>
<td>51-60</td>
<td>6.7</td>
<td>84</td>
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<td>61-70</td>
<td>4.2</td>
<td>82</td>
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<tr>
<td>71-80</td>
<td>3.9</td>
<td>80</td>
</tr>
<tr>
<td>81-90</td>
<td>2.1</td>
<td>76</td>
</tr>
<tr>
<td>91-100</td>
<td>1.6</td>
<td>73</td>
</tr>
<tr>
<td>101-110</td>
<td>1.4</td>
<td>42</td>
</tr>
<tr>
<td>111-120</td>
<td>1.2</td>
<td>36</td>
</tr>
<tr>
<td>1-120</td>
<td>6.83</td>
<td>64.5</td>
</tr>
</tbody>
</table>

Table 2: Lesion and calcification decomposition at bin width of 10keV

.3 Appendix C

The experiment performed at bin width of 5keV during lesion and calcification decomposition are presented here.
<table>
<thead>
<tr>
<th>Energy Bin (keV)</th>
<th>Lesion contrast (percent)</th>
<th>Calcification peak contrast (percent)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-5</td>
<td>30.2</td>
<td>91</td>
</tr>
<tr>
<td>6-10</td>
<td>28.3</td>
<td>89</td>
</tr>
<tr>
<td>11-15</td>
<td>14.1</td>
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<td>12.5</td>
<td>87</td>
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<td>101-105</td>
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<td>111-115</td>
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<td>35</td>
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<tr>
<td>116-120</td>
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<td>33</td>
</tr>
<tr>
<td>1-120</td>
<td>6.83</td>
<td>64.5</td>
</tr>
</tbody>
</table>

Table 3: Lesion and calcification decomposition at bin width of 5keV
.4 Appendix D

Some reconstructed images and sinograms during the process of spectral decomposition for both lesion and calcification are presented here.

Figure 1: Reconstruction with [a] 20keV bin width at (21-40)keV, [b] 10keV bin width at (101-110)keV, [c] with 5keV bin width at (111-115)keV, [d], with 5keV bin width at (101-105)keV
Figure 2: Sinograms generated during the process of decomposition [a] 20keV bin width at (21-40)keV, [b] 10keV bin width at (101-110)keV, [c] with 5keV bin width at (111-115)keV, [d], with 5keV bin width at (101-105)keV
.5 Appendix E
Figure 3: Code generated for the experiments
%Dose code
step=10;
count1=1;
for ii=1:step:120
seg = mono_proj(:,:,ii:ii+step-1);
total_photons = (size(seg,1)*size(seg,2)*size(seg,3));
extit_photons = sum(sum(seg));
photons_absorbed(count1,1)=ii;
photons_absorbed(count1,2) = (1-exit_photons/total_photons)*100;
count1=count1+1;
x=0;
y=0;
for jj=1:size(seg,3)
    x=x+mean((ii+jj-1)*seg(:,:,jj));
    y=y+(ii+jj-1);
end
x=x/step;
y=y/step;
average_energy(count1,1)=y;
average_energy(count1,2)=x;
end
disp(poly_proj);
count=count+1;
end
disp(photons_absorbed);

Figure 4: Code used in generating the absorbed dose at different energy bins
Bibliography


[7] Hyun Woo Chung, Young So, Jung-Hyun Yang, Kyoung Sik Park, Young Bum Yoo, Nami Choi, Mi Young Kim, Jayoun Kim, and Eun Jeong Lee. Adjunctive


[45] Etta D Pisano, Jayanthi Chandramouli, Bradley M Hemminger, Deb Glueck, R Eugene Johnston, Keith Muller, M Patricia Braeuning, Derek Puff, William


