X-RAY IMAGING OF NON ACTIVE MEDICAL IMPLANTS

PhD THESIS

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X-RAY IMAGING OF NON ACTIVE MEDICAL IMPLANTS

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Αφιερώνω αυτήν την διατριβή στην οικογένειά μου, στους αγαπημένους πατέρα, μητέρα και αδερφό μου.
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## TABLE OF CONTENTS

Table of contents................................................................. 7

1. **INTRODUCTION** ................................................................. 16
   1.1. Motivation........................................................................ 16
   1.2. Thesis outline............................................................... 16

2. **PRODUCTION OF X-RAYS AND INTERACTIONS WITH MATTER** ............... 18
   2.1. Bremsstrahlung radiation............................................ 19
   2.2. Characteristic radiation............................................... 20
   2.3. Synchrotron Radiation................................................ 21
   2.4. Factors controlling the x-ray beam............................. 22
   2.5. Interactions of x-rays with matter............................... 25
   2.6. Beam attenuation and absorption............................... 28
   2.7. Dosimetry................................................................. 29

3. **BREAST IMAGING** .............................................................. 32
   3.1. Anatomy of the breast................................................ 32
   3.2. Mammography.......................................................... 34
   3.3. Breast abnormalities.................................................. 35
   3.4. Breast Tomomosynthesis........................................... 38
     3.4.1. Principles of tomosynthesis.................................... 38
     3.4.2. Systems and acquisition technologies................... 41
     3.4.3. Reconstruction algorithms................................. 42
     3.4.4. Evaluation metrics................................................ 44

4. **BREAST AUGMENTATION** .................................................. 46
   4.1. Breast implants.......................................................... 47
   4.2. Incision and implant placement................................. 48
4.3. Breast Implant and Mammography

4.4. Physical Properties of Silicone Gel

REFERENCES PART 1

5. EVALUATION OF THE EFFECT OF SILICONE BREAST INSERTS ON MAMMOGRAPHY AND BREAST TOMOSYNTHESIS IMAGES: A Monte Carlo Simulation Study

5.1. Introduction

5.2. Software phantoms

5.3. X-ray imaging simulation

5.3.1. Simulated Imaging protocols

5.3.2. Optimal Parameters for BT acquisition arcs

5.3.3. Monte Carlo code

5.3.4. Image quality evaluation metrics

5.4. Results in the case of mammographic images

5.4.1. Step-wedge phantom

5.4.2. Snail phantom

5.4.3. Phantom with a realistic silicone gel implant shape

5.5. Results in the case of tomosynthetic images

5.5.1. Step-wedge phantom

5.5.2. Snail phantom

5.5.3. Phantom with a realistic shape of silicone gel implant

5.6. Performance of Mammography versus BT

5.7. Conclusion

6. COMPEX BREAST PHANTOM WITH SILICONE GEL

6.1. Complex phantom

6.2. Mammographic examination

6.3. Tomosynthesis examination

6.4. Further Investigation
7. INVESTIGATION OF ALTERNATIVE IMAGING MODALITIES ....................................................... 90

7.1. Dual Energy subtraction mammography .............................................................................. 91

7.2. Dual Energy subtraction BT ............................................................................................... 94

7.3. DE with the use of breast phantom .................................................................................... 96

7.4. BT performed in a discontinuous acquisition arc .............................................................. 99

7.5. Conclusion ......................................................................................................................... 100

8. INVESTIGATING PHASE CONTRAST AS A PROMISING TECHNIQUE FOR SILICONE EFFECTS LIMITATION ......................................................................................................................... 102

8.1. Phase contrast simulation study ......................................................................................... 105

9. IMAGE QUALITY EVALUATION OF DIGITAL MAMMOGRAPHY TECHNIQUES WITH PHASE CONTRAST SYNCHROTRON RADIATION ............................................................................................................... 109

9.1. Introduction .......................................................................................................................... 109

9.2. Hardware Phantom ............................................................................................................. 110

9.3. Image acquisition and reconstruction process ..................................................................... 111

9.4. Evaluation metrics ............................................................................................................. 114

9.5. Results ............................................................................................................................... 116

9.6. Discussion .......................................................................................................................... 123

9.7. Conclusion ......................................................................................................................... 125

10. CONCLUSION ...................................................................................................................... 127

REFERENCES PART 2 .................................................................................................................. 131
Publications

Publications in international peer-reviewed journals:


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- Daskalaki A, Bliznakova K & Pallikarakis N. *Software Breast Phantoms for Phase Contrast Imaging Applications*, Toronto 2015
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ABSTRACT

The current thesis is focused on breast implants and their imaging with the use of x-rays. During the last years the number of breast augmentations with silicone implants is increasing, especially in America. More and more women want to have breast implants either for esthetic reasons or for physiological reasons after mastectomy. On the other hand, several characteristics of silicone gel implants and the techniques of their placement affect the x-ray based imaging of the breast. In addition, the presence of silicone gel-filled implants interferes with standard mammography, since silicone is a radiopaque material.

The aim of this research is to explore the effects of silicone gel implant insertion in breasts and how the thickness of an implant as well as the incident beam energy interfere with the detectability of high and low contrast lesions. This study also aims in the investigation of non-conventional imaging techniques in case of augmented breasts in order to improve the detectability of lesions and the quality of breast images.

For the purpose of this study a wide number of software phantoms with homogeneous and heterogeneous texture were constructed for simulation studies with the use of the XRayImagingSimulator. A breast hardware phantom consisting of materials mimicking breast lesions was also constructed and used at the ELETTRA synchrotron facilities in Trieste, Italy for experiments performed with synchrotron radiation. Mammography and breast tomosynthesis were investigated as conventional breast imaging modalities in the incident beam energy range 20 KeV to 30 keV. Moreover, alternative imaging modalities like Dual Energy subtraction mammography, Dual Energy subtraction breast tomosynthesis, tomosynthesis with a discontinuous acquisition arc and In-Line phase contrast imaging were studied as potential modalities for augmented breast imaging.

This study established upper silicone gel thickness limits in the conventional beam energy range for both mammography and tomosynthesis above which microcalcifications could not be detected. Comparisons between tomosynthetic and mammographic images showed that the latter have superior CNR values compared to those calculated for tomosynthesis. Alterations of energy in the interval of 20 keV to 30 keV led to improved detectability of microcalcifications overlapped by silicone, while the opposite stands for breast masses. Although tomosynthesis resulted in inferior image quality in terms of CNR, it demonstrated an advantage in visualizing a larger breast area and small low contrast lesions because of non-overlaps. Moreover it was shown that DE mammography improves the detectability of high contrast features overlapping with high silicone gel thicknesses but gives marginal CNR values. Dual Energy
breast tomosynthesis resulted in a superior image quality compared with the rest of the imaging modalities studied, when the visualization of both high and low contrast lesions in the neighborhood of silicone is needed. Breast tomosynthesis with an alternative acquisition arc succeeded to visualize microcalcifications under high thicknesses of silicone gel with sufficiently high CNR. This indicates that tomosynthesis with an appropriate acquisition arc might resolve the lack of contrast appearing in breast images with silicone gel insertion. Finally, simulations as well as experimental studies showed that phase contrast imaging improves the image quality resulting in higher CNR values and in a strong edge enhancement of features, which may be useful in augmented breasts imaging.
ΠΕΡΙΛΗΨΗ

Η παρούσα διατριβή εστιάζεται στα εμφυτεύματα στήθους και στον τρόπο με τον οποίο απεικονίζονται με τη χρήση ακτίνων-χ. Τα τελευταία χρόνια, οι προσθετικές στήθους με χρήση εμφυτευμάτων σιλικόνης αυξάνονται. Ωλοένα και περισσότερες γυναίκες επιθυμούν να προβούν σε μία πλαστική στήθους είτε για λόγους αισθητικούς, είτε για λόγους φυσιολογικούς, όπως στην περίπτωση μαστεκτομής. Από την άλλη πλευρά, αρκετά χαρακτηριστικά αυτών των εμφυτευμάτων σιλικόνης καθώς και οι τεχνικές τοποθέτησής τους επηρεάζουν την απεικόνιση του στήθους με χρήση ακτίνων-χ. Επίσης, η παρουσία των εμφυτευμάτων σιλικόνης παρεμποδίζει την μαστογραφική απεικόνιση, αφού η σιλικόνη είναι ακτινοσκιερό υλικό.

Στόχος της έρευνας αυτής είναι η διερεύνηση των συνεπειών ενός εμφυτεύματος σιλικόνης στις καθημερινές τεχνικές απεικόνισης του μαστού, αλλά και του τρόπου με τον οποίο το πάχος του εμφυτεύματος καθώς και η ενέργεια της προσπίπτουσας ακτινοβολίας επηρεάζουν την ανίχνευση δομών χαμηλής και υψηλής αντίθεσης. Επιπλέον, η μελέτη αυτή στοχεύει στην εξέταση μη συμβατικών τεχνικών απεικόνισης, με στόχο τη βελτίωση της ανίχνευσης αλλοιώσεων του μαστού και της ποιότητας απεικόνισης του μαστού με εμφύτευμα σιλικόνης.

Για τις ανάγκες αυτής της μελέτης κατασκευάστηκαν ομοιώματα με ομοιογενή και ανομοιογενή σύσταση για μελέτες προσομοίωσης με τη χρήση του λογισμικού XRayImagingSimulator. Επίσης, ένα ομοίωμα μαστού το οποίο εμπεριείχε υλικά μιμούμενα αλλοιώσεις του μαστού, κατασκευάστηκε για πειραματικές μελέτες στις εγκαταστάσεις σύγχροτου έργου ELETTRA, στην Teorhētēs tis Italías. Η μαστογραφία και η τομοσύνθεση διερευνήθηκαν ως κλασικές μέθοδοι απεικόνισης μαστού, με χρήση μονοενεργειακής ακτινοβολίας στο διάστημα 20-30 keV με 30 keV. Επιπλέον, εναλλακτικές μέθοδοι απεικόνισης όπως η διπλής ενέργειας μαστογραφία, η διπλής ενέργειας τομοσύνθεση μαστού, η τομοσύνθεση διπλού τόξου και η απεικόνιση αλλαγής φάσης μελετήθηκαν ως εν δυνάμει τεχνικές για την απεικόνιση μαστού με εμφύτευμα σιλικόνης.

Μέσω αυτής της μελέτης εδραιώθηκαν τα άνω όρια πάχους σιλικόνης, εκτός των οποίων η ανίχνευση μικροαποτιτανώσεων είναι αδύνατη τόσο μέσω μαστογραφίας όσο και μέσω τομοσύνθεσης στις συμβατικές ενεργειακές ώρες απεικόνισης μαστού. Συγκρίσεις μεταξύ των μαστογραφικών εικόνων και των εικόνων τομοσύνθεσης εδειξαν ότι οι πρώτες χαρακτηρίζονται από υψηλότερη αντίθεση. Εναλλαγές της ενέργειας εντός του διαστήματος 20-30 keV οδήγησαν σε βελτιωμένη ανίχνευση μικροαποτιτανώσεων οι οποίες υπερκαλύπτονταν από σιλικόνη, ενώ για χαμηλής αντίθεσης μάζες
διαπιστώθηκε το αντίθετο. Αν και η τομοσόγνθηση κατέληξε σε χαμηλότερη ποιότητα εικόνας όσον αφορά την αντίθεση, παρουσίασε πλεονεκτήματα στην απεικόνιση μεγαλύτερης περιοχής μαστογραφικού ενδιαφέροντος και μικρών μαζών χαμηλής αντίθεσης μετριάζοντας το πρόβλημα της επικάλυψης. Επίσης απεδείχθη ότι η μαστογραφία διπλής ενέργειας βελτιώνει την ανίχνευση δομών υψηλής αντίθεσης, οι οποίες υπερκαλύπτονται από σιλικόνη μεγάλου πάχους αλλά με οριακές τιμές του λόγου αντίθεσης/θορύβου. Τα αποτελέσματα της τομοσόγνθησης μαστού διπλής ενέργειας παρουσίασαν βελτωμένη ποιότητα εικόνας συγκρινόμενα με τα αποτελέσματα των υπόλοιπων τεχνικών απεικόνισης, στην περίπτωση όπου απαιτείται η ανίχνευση αλλοίωσεων υψηλής και χαμηλής αντίθεσης. Η τομοσόγνθηση μαστού με χρήση εναλλακτικού τόξου λήψης πέτυχε να απεικονίσει με αρκετά υψηλή αντίθεση, μικροαποτιτανώσεις επικαλυπτόμενες από σιλικόνη μεγάλου πάχους. Το παραπάνω αποτέλεσμα υποδεικνύει ότι η τομοσόγνθηση με χρήση κατάλληλου τόξου λήψης μπορεί να επιλύσει το πρόβλημα της έλλειψης αντίθεσης που παρουσιάζεται στις απεικόνισεις μαστού με εμφύτευμα σιλικόνης. Τέλος, προσομοιώσεις καθώς και πειραματικές μελέτες έδειξαν ότι η απεικόνιση με χρήση αντίθεσης φάσης βελτιώνει την ποιότητα της εικόνας οδηγώντας σε υψηλότερες τιμές του λόγου αντίθεσης/θορύβου και σε έντονη ενίσχυση στις ακμές δομών, στοιχεία τα οποία μπορούν να βελτιώσουν την απεικόνιση στήθους με εμφύτευμα σιλικόνης.
In the first part, the theoretical background of this thesis is presented. An introduction to x-ray radiation is made and the way it interacts with matter is explained. The anatomy of the breast and the principles of breast imaging are analyzed. Breast tomosynthesis and the main reconstruction algorithms and acquisitions used are introduced. Finally, the types of breast implants as well as the effects that come along with their insertion in breast tissue are presented.
1. INTRODUCTION

1.1. Motivation

Cancer incidence and mortality statistics show that breast cancer is one of the most common types of cancer worldwide even nowadays. Despite the technical improvements in breast imaging, early diagnosis and lower rate of missed lesions remain a primary goal against breast cancer. Studies have shown that breast implants, especially silicone gel implants, contribute in an opposite direction. Breast augmentation has become one of the most popular cosmetic surgeries worldwide, whereas the number of silicone gel implantations into breasts is continuously increasing. The fact that the visibility of lesions is of a great importance in breast imaging and that a breast implant may interfere in their detectability acted as a trigger for this study. The aim of this thesis is to investigate the effects of breast implant insertion on the detectability and visibility of lesions on conventional breast imaging techniques, as well as to provide new data and further knowledge on the limitations of breasts imaging under this circumstances. Moreover, this study intends to provide alternative techniques and suggestions in order to improve the image quality of a mammogram in case of silicone gel implantation.

1.2. Thesis outline

This thesis is organized in two Parts with 10 Chapters in total. Part 1 includes the theoretical study of the thesis and is composed of 4 chapters. Firstly, an introduction with the aim of this study is presented. Chapter 2 summarizes the main characteristics of x-ray radiation, including the way of its production and its interactions with matter. The anatomy of the breast and the main principles of breast imaging are described in Chapter 3, whereas an extended review on mammography and breast tomosynthesis is also presented. Chapter 4, the last of Part 1, is dedicated to breast implants. The types of implants and their way of placement into breasts are presented. Moreover, the characteristics of silicone gel implants and their way of interaction with mammography are described in detail.

Part 2 contains the simulated and experimental studies of this thesis along with their results and discussions. Specifically, in Chapter 5 an extended work on the imaging drawbacks of silicone gel insertion into breasts in mammography and breast tomosynthesis is presented. The projection images studied are acquired through simulations with the use of three software phantoms. Also the detectability of high and low contrast lesions in the presence of silicone gel of different thicknesses is investigated.
Chapter 6 is focused on the imaging drawbacks of silicone gel into a simulated complex breast phantom under compression. In Chapter 7 alternative imaging modalities are studied as possible techniques for improved image quality of augmented breasts. The modalities under consideration were: Dual Energy subtraction mammography, Dual Energy breast tomosynthesis and breast tomosynthesis in a discontinuous arc. The advantages and disadvantages of these modalities in detecting and visualizing high and low contrast lesions in the presence of silicone are presented. Chapters 8 and 9 are dedicated to phase contrast imaging. Chapter 8 demonstrates a simulated work, pointing out that phase contrast imaging can improve the image quality of augmented breasts. Chapter 9 is dedicated to an experimental study of phase contrast breast imaging accomplished at ELETTRA synchrotron facilities in Trieste. Projection images acquired with the use of a hardware phantom mimicking the breast in both mammography and tomosynthesis mode are analyzed. Finally, Chapter 10 summarizes the results and the conclusions of the present thesis and points out future research directions in this field.
2. PRODUCTION OF X-RAYS AND INTERACTIONS WITH MATTER

Over a century ago, in 1895, Wilhelm Conrad Roentgen discovered that a high voltage discharge between electrodes in gas at very low pressure produces a penetrating radiation which causes certain material to fluorescence visible light (Roetgen 1895). He observed that if a voltage exceeds about 30 kV, then the radiation - which he called x-rays - can penetrate a hand, casting shadows of the bones on a fluorescent screen. It was soon understood that electrons, emitted from the negative electrode (cathode) of the discharge tube and accelerated by the high applied voltage, emit electromagnetic radiation when they collide with the positive electrode (anode). Roentgen's work showed how to produce electromagnetic radiation with wavelengths comparable to atomic dimensions. As a consequence, x-rays proved to be a powerful means for physicists to probe the structure of atoms and for medical doctors to probe the human body, diagnose and treat diseases.

Fig. 2.1: X-ray tubes are illustrated: (a) early gas x-ray tubes, (b) a schematic diagram of x-ray tube from William Coolidge's 1913 patent application and (c) a vacuum x-ray tube of the type used nowadays.

The key to Roentgens discovery was a device called a Crooke’s tube, which was a glass envelope under high vacuum, with a wire element at one end forming the cathode, and a heavy copper target at the other end, forming the anode (Fig. 2.1). When a high voltage was applied to the electrodes, electrons formed at the cathode would be pulled towards the anode and strike the copper with high energy. An x-ray tube and an x-ray generator are the necessary components for x-ray production and control. The x-ray tube provides the proper environment and components to produce x-rays, whereas the x-ray generator provides the source of electrical voltage controllable by the user. The basic components of an x-ray tube are the
two electrodes, the cathode and the anode, positioned in a small distance apart (1-2 cm) in a vacuum enclosure called the insert, made of either glass or metal. The cathode is the negative terminal of the x-ray tube. It is a filamentous structure and when current is flown through it, the filament gets heated and starts emitting electrons on its surface by the process called thermionic emission. High voltage is applied between the cathode and the anode. High voltage of kilovolt range (50-150 kV) supplied by the x-ray generator causes electrons to move towards the positive terminal of the tube at a velocity of half the velocity of light. The anode is the positive terminal of the tube, generally made of tungsten (W, Z=74) disc. Other materials used are molybdenum (Mo, Z=42) and rhodium (Rh, Z=45), both famous in mammography x-ray tubes. Fast moving electrons traveling from the filament (cathode) to the target (anode) convert a small percentage (1%) of their kinetic energy into x-ray photons by the formation of bremsstrahlung and characteristic radiation.

2.1. Bremsstrahlung radiation

Bremsstrahlung interactions, the primary source of x-ray photons from an x-ray tube, are produced by the sudden stopping, breaking or slowing of high-speed electrons at the target. An electron comes in close proximity to the nucleus of a target atom and experiences attractive forces due to the positive charge of the protons in the nucleus. This combined positive charge decelerates and changes the direction of the electron, the magnitude of which is inversely proportional to the impact parameter distance, i.e. 1/distance (Bushberg et al 2002). The kinetic energy lost is converted to electromagnetic radiation with equivalent energy in a process known as bremsstrahlung (a German term meaning “braking radiation”), as shown in Fig. 2.2.

When the electrons from the filament strike the tungsten target, x-ray photons are created if they either hit a target nucleus directly (rare) or their path takes them close to the nucleus. If a high-speed electron hits the nucleus of a target atom, all its kinetic energy is transformed into a single x-ray photon. In other words, total absorption has occurred. Thus, the energy of the resultant photon (keV) is numerically equal to the energy of the electron. This in turn is equal to the kilovoltage applied across the x-ray tube at the instant of its passage. This happens rarely since the probability of interaction decreases as the distance of interaction decreases. More frequently high-speed electrons pass through the close neighborhood of the positively charged nucleus experiencing electrostatic attraction forcing them to decelerate. This deceleration of electrons and loss of kinetic energy is converted to photons. The closer the electrons pass to the nucleus the greater is the loss of their energy and the resulting energy of the photon. As a result, the
output is a continuous spectrum of x-ray energies with maximum x-ray energy (in keV) determined by the peak potential difference (in kVp) as shown in Fig. 2.3.

![Fig. 2.2: X-ray production by energy conversion. Events 1, 2 and 3 depict bremsstrahlung radiation, whereas event 4 demonstrates the production of characteristic radiation.](image)

A larger number of low energy x-rays are produced in the output spectrum, simply due to the lower probability of interaction closer to the nucleus. Other reasons for the continuous spectrum of energy are the varying voltage difference between the target and the filament as well as the fact that most electrons participate in many interactions before all the kinetic energy is expended.

### 2.2. Characteristic radiation

Characteristic radiation occurs when an electron from the filament displaces an electron from an inner-shell of the tungsten target atom, thereby ionizing the atom (Fig. 2.2). When this happens, another electron in an outer-shell of the tungsten atom is quickly attracted into the void in the deficient inner-shell. When the displaced electron is replaced by the outer-shell electron, a photon is emitted with an energy equivalent to the difference in the two orbital binding energies. Characteristic radiation from the K-shell occurs only above 70 kVp with a tungsten target and occurs as discrete increments compared with Bremsstrahlung radiation. In Figure 3 the radiation spectra for tungsten at four different x-ray tube kVps are illustrated, with the continuous curves corresponding to bremsstrahlung radiation and the peaks to characteristic radiation of tungsten.
The energies of characteristic photons are a function of the energy levels of various electron orbital levels and hence are characteristic of the target atoms. Characteristic radiation has a higher intensity, is preferred but is only a minor source of radiation from an x-ray tube.

2.3. Synchrotron Radiation

A third mechanism by which x-rays are produced is synchrotron emission. Synchrotron radiation is emitted by charged particles accelerated in a curved path or orbit. Classically, any charged particle which moves in a curved path or is accelerated in a straight-line path will emit electromagnetic radiation. Synchrotrons are circular (ring-shaped) constructions that are able to produce electromagnetic radiation of great brightness (high intensity) and of very wide range. The basic principle of operation is the acceleration of electron in a ring-shaped vacuum tube, using electromagnetic field. The high-speed electron beam is forced to change direction using magnets, thus the electrons emit electromagnetic radiation. A schematic of a synchrotron can be seen in Fig. 2.4.
Electrons enter the ring already accelerated using a linear accelerator (LINAC). Once inside the synchrotrons they can be further accelerated (thus acquiring more energy) by the application of radio frequency electric fields. At certain points (where the ring has to turn) a bending magnet is placed. Bending magnets force the electrons to change their direction and as a result they emit electromagnetic radiation.

Fig. 2.4: Schematic view of synchrotron

This, however, is not the only way that electromagnetic radiation is generated in a synchrotron facility. New generation synchrotrons use undulators, which are able to produce much brighter x-ray beams than bending magnets. In an undulator there are a series of dipole magnets placed in determined distance (λ) in arrays. When the electrons pass through this high alternating magnetic field, they are forced to oscillate and thus they emit x-ray radiation. The beam resulting from an undulator is highly polarized and it can be controlled depending on the magnet structure, from linear to circular. Synchrotrons can produce a very broad energy range from infrared to hard x-rays. The main advantage of synchrotrons is that they can produce very “high quality” beams in terms of: very high intensity (more than a million times higher than that of conventional x-ray tubes), very broad and continuous spectra (and monochromatic beams), highly collimated and polarized beams.

2.4. Factors controlling the x-ray beam

The x-ray beam emitted from an x-ray tube may be modified to suit the needs of the application by altering the beam exposure length (timer), exposure rate (mA), beam energy (kVp and filtration), beam shape (collimation), and source-to-patient distance (long or short cone).
**Exposure Time:** Portrays the changes in the x-ray spectrum that occur when the exposure time is increased while the tube current (mA) and voltage (kVp) remain constant. When the exposure time is doubled, the number of photons generated is doubled, but the range intensity of photons energies is unchanged. Therefore, changing the time simply controls the “quantity” of the exposure, i.e. the number of photons generated. The amount of radiation that a patient receives is determined by the mAs (mA x time).

**Tube Current (mA):** Illustrates the changes in the spectrum of photons that result from increasing tube current (mA) while maintaining constant tube voltage (kVp) and exposure time. As the mA setting is increased, more power is applied to the filament, which heats up and releases more electrons that collide with the target. A linear relationship exists between mA and radiation output. The quantity of radiation produced (mAs) is expressed as the product of time and tube current. The quantity of radiation remains constant regardless of variations in mA and time as long as their product remains constant.

**Tube Voltage (kVp):** Increasing the kVp increases the potential difference between the cathode and anode, thus increasing the energy of each electron when it strikes the target. The greater the potential difference the faster the electrons travel from the cathode to the anode. This results in an improved efficiency of conversion of electron energy into x-ray photons, and thus an increase in the numbers of photons generated, in their mean energy and maximal energy.

The increased number of high-energy photons produced per unit time by use of higher kVp results from the greater efficiency in the production of bremsstrahlung photons that occurs when an increased number of higher-energy electrons interact with the target.

The ability of x-ray photons to penetrate matter depends on their energy. High-energy x-ray photons have a greater probability of penetrating matter, whereas relatively low-energy photons have a greater probability of being absorbed. Therefore, the higher the kVp and mean energy of the x-ray beam, the greater the penetrability of the beam through matter and the lower the radiation that is absorbed by the patient.

**Half Value Layer:** A useful way to characterize the penetrating quality of an x-ray beam is its half-value layer (HVL). The HVL is the thickness of an absorber, such as aluminium, required to reduce by one half the number of x-ray photons passing through it. As the average energy of an x-ray beam increases, so does its HVL. The term quality refers to the mean energy of an x-ray beam. Half value layer measures the intensity of a beam.
Filtration: An x-ray beam consists of a spectrum of x-ray photons of different energies, but only photons with sufficient energy to penetrate anatomic structures and reach the image receptor are useful for diagnostic radiology. Those that are of low energy (long wavelength) contribute to patient exposure but do not have enough energy to reach the detector. The higher the kVp, the less radiation is absorbed by the patient. Consequently, to reduce patient dose, the less penetrating photons should be removed. This can be accomplished by placing an aluminium filter in the path of the beam. The aluminium presents the advantage of removing many of the lower-energy photons with lesser effect on the higher energy photons that are able to penetrate the film.

Collimation: A collimator is a metallic barrier with an aperture in the middle used to reduce the size and shape of the x-ray beam and therefore the volume of irradiated tissue within the patient. The round collimator is a thick plate of radiopaque material (usually lead) with a circular opening centered over the port in the x-ray through which the x-ray beam emerges. Typically, round collimators are built into open-ended aiming cylinders. Rectangular collimators further limit the beam to a size just larger than that of the x-ray film. The size of the beam should be reduced to the size of the film being exposed to reduce further unnecessary patient exposure. Some types of film-holding instruments also provide rectangular collimation of the x-ray beam.

The use of collimation also improves image quality. When an x-ray beam is directed at a patient, about 90% of the x-ray photons are absorbed by the tissues and 10% of the photons pass through the patient and reach the film. Many of the absorbed photons generate scattered radiation within the exposed tissues by a process called Compton scattering. These scattered photons travel in all directions, many of which reach the film and thereby degrade image quality. The detrimental effect of scattered radiation of the images can be minimized by collimating the beam to reduce the number of scattered photons reaching the film.

Inverse Square Law: The intensity of an x-ray beam at a given point (number of photons per cross-sectional area per unit exposure time) depends on the distance of the measuring device from the focal spot. For a given beam the intensity is inversely proportional to the square of the distance from the source. The reason for this decrease in intensity is that the x-ray beam spreads out as it moves from the source. The relationship is as follows:

$$\frac{I_1}{I_2} = \frac{D_2^2}{D_1^2}$$

where D is the distance and I is the intensity. Therefore, changing the distance between the x-ray tube and the patient has a marked effect on beam intensity. Such a change requires a corresponding modification of the kVp or mAs if the exposure of the film is to be kept constant.
2.5. Interactions of x-rays with matter

The intensity of an x-ray beam is reduced by interaction with the matter it encounters. This attenuation results from interactions of individual photons in the beam with atoms in the absorber (patient). The x-ray photons are either absorbed or scattered out of the beam. In scattering, photons are ejected out of the primary beam as a result of interactions with the orbital electrons of absorber atoms. There are three mechanisms where these interactions take place: (1) coherent scattering, (2) Compton scattering, and (3) photoelectric absorption for the energy interval used in mammography. Of course there is also pair production that can only occur when the incident x-ray photon has energy greater than 1.02 MeV. In addition, about 9% of the primary photons pass through the patient without interaction to produce the image. All the mechanisms are presented in detail in Fig. 2.5.

![Fig. 2.5: Illustrative summary of x-ray interactions. (A) Primary, unattenuated beam does not interact with material. (B) Photoelectric absorption results in total removal of incident x-ray photon, with excess energy distributed to kinetic energy of photoelectron. (C) Coherent scattering with an electron or the whole atom in which no energy is exchanged. (D) Compton scattering interactions occur with essentially unbound electrons, with transfer of energy between recoil electron and scattered photon.](image)

**Coherent scattering**

Coherent scattering, also known as classical scattering and Thomson scattering, may occur when a low-energy incident photon, less than 10 keV, passes near an outer electron of an atom (which has a low binding energy). The incident photon interacts with the electron thus enabling the atom as a whole to take up the recoil, by causing it to vibrate momentarily at the same frequency as the incoming photon. The
incident photon then ceases to exist. The vibration causes the electron to radiate energy in the form of another x-ray photon with the same frequency and energy as in the incident photon but the direction of the incident x-ray photon is altered. This is the only interaction within the matter in which no energy is deposited in the scattering medium and no ionization occurs. The probability of coherent scattering is given by the symbol $\sigma_{coh}$. In soft tissue, probability of this event occurring is low, on the order of 5% of all scattering events, because of the low effective atomic number of soft tissues ($Z=7.5$). Coherent scattering varies with the atomic number of absorber ($Z$) and incident photon energy ($E$) by: $Z^2/E$.

**Compton scattering**

Compton scattering occurs when a photon interacts with an outer orbital electron, which receives kinetic energy and recoils from the point of impact. The incident photon is then deflected by its interaction and is scattered from the site of the collision. The energy of the scattered photon equals the energy of the incident photon minus the kinetic energy gained by the recoil electron plus its bonding energy. As with photoelectric absorption, Compton scattering results in the loss of an electron and ionization of the absorbing atom. The energy of the scattered photon is given by the equation known as Klein-Nishina:

$$h\nu_{sc.ph} = h\nu_{ph} \frac{1}{1+a(1-\cos\theta)}$$  

Equation 2.2

while the energy of the Compton electron is given by the equation:

$$E_e = h\nu_{ph} \frac{a(1-\cos\theta)}{1+a(1-\cos\theta)}$$  

Equation 2.3

where $h\nu_{sc.ph}$ and $h\nu_{ph}$ is the energy of the scattered and incoming photon respectively, $\theta$ is the angle of the scattered photon and $a = \frac{h}{m_0c^2}$; $m_0c^2$ is the rest mass of the electron and equals to 0.511 MeV and $h\nu$ is the incoming photon energy.

Scattered photons travel in all directions. The higher the energy of the incident photon, however, the greater the probability that the angle of scatter of the secondary photon will be small and its direction will be forward. On one hand, this is advantageous to the patient because some of the energy of the incident x-ray beam escapes the tissue, but, on the other hand, it is disadvantageous because it causes nonspecific film darkening (or fogging of the film). Scattered photons darken the film while carrying no useful information to it because their path is altered.

The probability of Compton scattering is given by the symbol $\sigma$ and is directly proportional to the electron and physical density of the material and inversely proportional to photon energy. The number of
electrons in bone is greater than in water, therefore the probability of Compton scattering is correspondingly greater in bone than in tissue. The importance of photoelectric absorption and Compton scattering in diagnostic radiography relates to differences in the way photons are absorbed by various anatomic structures. The number of photoelectric and Compton interactions is greater in hard tissues than in soft tissues. As a consequence, more photons in the beam exit the patient after passing through soft tissue than through hard tissue.

**Photoelectric absorption**

Photoelectric absorption occurs when an incident photon collides with an inner-shell electron in an atom of the absorbing medium resulting in total absorption and the incident photon ceases to exist. The electron is ejected from its shell, resulting in ionization and becomes a recoil electron (photoelectron). The kinetic energy imparted to the recoil electron is equal to the energy of the incident photon minus that used to overcome the binding energy of the electron. Most photoelectric interactions occur in the K shell because the density of the electron cloud is greater in this region and a higher probability of interaction exists. The energy of the photoelectron is given by the equation:

\[ E_e = h\nu_{ph} - E_{bind} \]  

Equation 2.4

An atom that has participated in photoelectric interaction is ionized. This electron deficiency (usually in the K shell) is instantly filled, usually by an L- or M-shell electron, with the release of characteristic radiation. The recoil electrons ejected during photoelectric absorptions travel only a short distance in the absorber before they give up their energy. As a consequence, all the energy of incident photons that undergo photoelectric interaction is deposited in the patient. This is beneficial in producing high-quality radiographs, because no scattered radiation fogs the film, but potentially deleterious for patients because of increased radiation absorption.

The probability of photoelectric absorption, commonly denoted as the symbol \( \tau \), is proportional to the cube of the atomic number of the interacting atom and inversely proportional to the cube of the incident photon energy, i.e.:

\[ \tau \propto \frac{Z^3}{(h\nu)^3} \]  

Equation 2.5

Photoelectric interaction is more likely to occur with higher atomic number elements and lower x-ray energies. These interaction dependencies have ramifications, such that larger x-ray transmission differences in patient anatomy occur at low energies, enhance signal contrast. This also means that
contrast agents, x-ray detectors, and protection devices are preferably made of high Z elements, such as iodine, gadolinium, and lead, respectively.

**Secondary electrons**

In both photoelectric absorption and Compton scattering, electrons are ejected from their orbits in the absorbing material after interaction with x-ray photons. These secondary electrons give up their energy in the absorber by either of two processes: (1) collisonal interaction with other electrons, resulting in ionization or excitation of the affected atom, and (2) radiative interactions, which produce bremsstrahlung radiation resulting in the emission of low-energy x-ray photons. Secondary electrons eventually dissipate all their energy, mostly as heat by collisional interaction, and come to rest.

### 2.6. Beam attenuation and absorption

As an x-ray beam travels through matter, individual photons are removed, primarily through photoelectric and Compton interaction in a process known as attenuation. The reduction of beam intensity is predictable because it depends on physical characteristics of the beam and absorber. The **linear attenuation coefficient** ($\mu$) is the fraction of photons interacting per 1-unit thickness of the material generally expressed in cm$^{-1}$. Linear attenuation coefficient values indicate the rate at which photons interact as they move through material and are inversely related to the average distance photons travel before interacting. The rate at which photons interact (attenuation coefficient value) is determined by the energy of the individual photons, the atomic number and density of the material.

Basically, when a narrow beam of mono-energetic photons, such as x-ray, penetrates a surface of a material, it experiences a change in its intensity, owing to absorption or scattering that occur during the travel of the beam across the material. The intensity of the beam the long of an axis parallel to the direction that photons travel can be calculated (according to Lambert law) as:

$$ I = I_0 e^{-\mu x} \quad \text{Equation 2.6} $$

where $I_0$ is the initial x-ray beam intensity, $x$ the traveled distance and $\mu$ the total linear attenuation coefficient of the absorber given by the sum of the individual interactions probabilities:

$$ \mu = \tau + \sigma + \sigma_{coh} \quad \text{Equation 2.7} $$

The photoelectric effect dominates at low photon energies (< 26 keV) in soft tissue. With higher energy photons at low Z materials, Compton scattering dominates. Rayleigh scattering comprises about 10% of
the interactions in mammography and 5% in chest radiography. The average distance traveled by a photon between two successive interactions is called *Mean Free Path (MFP)*, and can be calculated from linear attenuation coefficient: $\lambda = 1/\mu$.

Since a linear attenuation coefficient is dependent on the density of a material, *the mass attenuation coefficient* is often reported for convenience. Consider water for example. The linear attenuation for water vapor is much lower than it is for ice because the molecules are more spread out in vapor so the chance of a photon encounter with a water particle is less. Normalizing $\mu$ by dividing it by the density of the element or compound will produce a value that is constant for a particular element or compound. This constant ($\mu/\rho$) is known as the mass attenuation coefficient and has units of cm$^2$/g. The relationship between mass attenuation and linear attenuation coefficient is:

\[
\text{Mass Attenuation Coefficient} = \frac{\text{Linear Attenuation Coefficient (}\mu\text{)} }{\text{Density (}\rho\text{)}}.
\]

The spectrum of photon energies (as illustrated by the kVp setting) in an x-ray beam is wide. In such a heterogeneous beam the probability of absorption of individual photons depends on their energy. Low-energy photons are much more likely than high-energy photons to be absorbed. As a consequence, the superficial layers of an absorber tend to remove the low energy photons and transmit the higher energy photons. Therefore, as an x-ray beam passes through matter, the intensity of the beam decreases but the mean energy of the resultant beam increases. This mechanism called *beam hardening*. In contrast to the absorption of a monochromatic beam, an x-ray beam is absorbed less and less by each succeeding unit of absorber thickness. In general, as the energy of the beam increases, so does the transmission of the beam through the absorber. However, when the energy of the incident photon is raised to the binding energy of the K-shell electrons of the absorber, the probability of photoelectric absorption increases sharply and the number of transmitted photons is greatly decreased. This is called k-edge absorption. (The probability that a photon will interact with an orbital electron is greatest when the energy of the photon equals the binding energy of the electron; on the contrary, it decreases as the photon energy increases.) Photons with energy less than the binding energy of K-shell electrons interact photoelectrically only with electrons in the L shell and in shells even farther from the nucleus.

2.7. Dosimetry

There are many different quantities and units used to quantify radiation, because there are a number of different aspects of an x-ray beam that can be used to express the amount of radiation. The selection of the most appropriate quantity depends on the specific application. Determining the quantity of radiation
exposure or dose is termed dosimetry. The term dose is used to describe the amount of energy absorbed per unit mass at a site of interest. Exposure is a measure of radiation based on its ability to produce ionization in air under standard conditions of temperature and pressure.

**EXPOSURE**

Exposure is the quantity most commonly used to express the amount of radiation delivered to a point. The conventional unit for exposure is the roentgen (R), and the SI unit is the coulomb per kilogram of air (C/kg): 1 R = 2.58 x 10^-4 C/kg and 1 C/kg= 3876 R. The reason exposure is such a widely used radiation quantity is that it can be readily measured. All forms of radiation measurement are based on an effect produced when the radiation interacts with a material. The specific effect used to measure exposure is the ionization in air produced by the radiation. The roentgen is the traditional unit of radiation exposure measured in air, 1 R is that amount of x-radiation that produces 2.08 x 10^9 ion pairs in 1 cc of air (STP). It measures the intensity of radiation to which an object is exposed. The roentgen applies only for x-rays and gamma rays.

In recent years the roentgen has been replaced by air kerma, acronym for kinetic energy released per unit mass (of air). Air kerma is another radiation quantity that is sometimes used to express the radiation concentration delivered to a point, such as the entrance surface of a patient's body. The typical unit for Kerma is the gray (Gy) that equals to J/kg. Moreover, for conversion between systems of units the following conversion formula is used: 1Gy=1J/kg=100rads.

**ABSORBED DOSE**

A human body absorbs most of the radiation energy delivered to it. The portion of an x-ray beam that is absorbed depends on the penetrating ability of the radiation and the size and density of the body section exposed. In most clinical situations more than 90% is absorbed. Absorbed dose is a measure of the energy absorbed by any type of ionizing radiation per unit mass of body tissue. Since an x-ray beam is attenuated by absorption as it passes through the body, all tissues within the beam will not absorb the same dose. The absorbed dose will be much greater for the tissues near the entrance surface than for those deeper within the body. The SI unit is the gray (Gy) – 1 Gy equals 1 joule/kg. The traditional unit of absorbed dose is the rad (radiation absorbed dose), where 1 rad is equivalent to 100 ergs/g of absorber, i.e. 1 Gy=100 rads.

**EQUIVALENT DOSE**

The equivalent dose (H) is a quantity used to measure the transfer of radiation (deposit) into matter per unit mass. But the transfer of energy and its deposit to matter can occur in different ways. The biological
effect because of absorbed dose depends on a) the dose rate (rate of depositing energy) and b) the density of ionization along the radiation path which is determined by the Linear Energy Transfer (LET). The LET is actually defined by the loss of energy through ionization per unit length. In order to take into account the different effects on biological tissues a quality factor is introduced (Q). The quality factor depends on the LET in a way that high LET radiations have high Q. The Q factor also depends on the radiation type and energy. The unit of equivalent dose is Sievert (Sv) and can be calculated as:

\[ H = \sum R Q \times D \quad \text{Equation 2.8} \]

where D is the absorbed dose. For diagnostic x-ray examinations, 1 Sv equals 1 Gy. Dose equivalent values can be converted from one system of units to the other by:

1 Sv = 100 rem

**EFFECTIVE DOSE**

As the equivalent dose takes into account the type and the energy of the radiation through the Q factor, the effective dose (E) takes into account the stochastic health risk. Not all human tissues react in the same way to ionizing radiation. The International Commission on Radiation Units and Measurements has assigned sensitivity factors to specified tissues and organs so that the effect of partial irradiation can be calculated if the irradiated regions are known (ICRU glossary). As a result, a tissue weighting factor, \( W_T \), is introduced that differs between body areas and tissues. Some examples of tissue weighting factors are: gonads 0.08, breast 0.12, lung 0.12, liver 0.04, oesophagus 0.4, thyroid 0.04, skin 0.01, and total body 1 (UNSCEAR 2008). The SI unit of effective dose is (Sv) and can be calculated as:

\[ E = \sum R W_T \times D \quad \text{Equation 2.9} \]

**MEAN GLANDULAR DOSE**

The Mean Glandular Dose (MGD) is a specialized quantity for measuring the dose in mammography representing the radiation risk for the breasts. The MGD is defined as the mean dose to glandular tissue based on the assumption that the part of breast that is more sensitive to radiation is the glandular tissue. In order to determine the MGD, the entrance surface exposure (air kerma) needs to be calculated first. The air kerma is usually calculated from the known calibration factors of the mammographic equipment. Furthermore, the breast surface exposure is translated into MGD by multiplying it with conversion
parameters depending on breast thickness, glandularity and incident spectra.

For comparison of imaging techniques, evaluation of equipment performance, general dose management, and regulatory and accreditation purposes, the MGD to a "standard" breast is used. The standard is a 4.2cm thick compressed breast consisting of 50% glandular and 50% adipose tissue. This corresponds to the standard phantom that is used for image quality evaluation and comparative dose determinations.

3. BREAST IMAGING

3.1. Anatomy of the breast

The mammary gland (called breast) is a complex organ that develops from the early weeks of gestation up to the start of lactation, when it becomes fully functional. Anatomically, the adult breast lies on the pectoralis major and extends usually from the second to the sixth rib. A part of the gland called the "axillary tail" extends towards the axilla along the lateral border of the pectoralis major. This is important because a mass of breast cancer can develop there, even though it might not seem to be part of the breast.

The breast, as described in Grey’s Anatomy (Bannister et al 1995), is composed of lobes which contain a network of glandular tissue consisting of branching ducts and terminal secretory lobules in a connective tissue stroma. Figure 3.1 depicts the anatomy of woman's breast. The lobules further consist of clusters of alveoli containing lactocytes (mammary secretory epithelial cells) that synthesize breast milk (Tobon and Salazar 1975). The alveoli are connected to small ducts that merge into larger ducts and finally end to one lactiferous duct. Thus, each lobe is drained by its own lactiferous duct having a tree-like structure. Then, each single lactiferous duct widens into a lactiferous sinus and narrows again terminating at the nipple. Fibrous strands consisting of condensations of connective tissue extend between the layer of deep fascia that covers the muscles of the anterior chest wall and the dermis. These suspensory ligaments (of Astley Cooper) are often well developed in the upper part of the breast and support the breast tissue, helping to maintain its non- ptotic form. Elsewhere in the normal breast, fibrous tissue surrounds the glandular components and extends to the skin and nipple, assisting the mechanical coherence of the gland. The interlobar stroma contains variable amounts of adipose tissue which is responsible for much of the increase in breast size at puberty.
Fig. 3.1: (A) Structure of the breast, (B) Changes in the breast during lactation, (C) Section of the nipple and (D) cross section of the nipple.

The breasts are supplied by branches of the axillary artery, the internal thoracic artery, and some intercostal arteries. The axillary artery supplies blood via the superior thoracic artery, the pectoral branches of the thoracoacromial artery, the lateral thoracic artery (via branches which curve around the lateral border of pectoralis major to supply the lateral aspect of the breast) and the subscapular artery. The internal thoracic artery supplies by 60% perforating branches to the anteromedial part of the breast. The second to fourth anterior intercostal arteries supply perforating branches more laterally in the anterior thorax: the second perforating artery is usually the largest, and supplies the upper region of the breast, and the nipple, areola and adjacent breast tissue (Cunningham 1977, Bannister et al 1995).
The lymph is drained from the breast by two main pathways. The majority of lymphatic vessels flow to the axillary nodes (70%), while the rest flow to internal mammary lymph nodes located deep within the breast. Knowledge of this lymphatic drainage is essential because in case of a breast cancer metastasis, the first lymph node of the chain is usually affected.

3.2. Mammography

Mammography is a specialized medical imaging technique that uses low-energy x-rays (usually around 30 kVp) to visualize the interior part of the breast. A mammography exam called mammogram aids in the early detection and diagnosis of breast malignancies at an early stage. During the procedure, the breast is under compression (120-150 N) between the detector and a compression radiotranslucent paddle. Compression reduces the thickness of the breast and as a result the scattered radiation and the dose delivered to breast tissue is less. Moreover, it causes much of the breast tissue to spread out, thus being of a uniform thickness, and eliminates motion blurring.

![Examples of variation in mammographic density](image)

The mammographic image depicts the result of the interactions between the breast tissue and the x-ray photons as described in the previous chapter. The x-ray photons reach the detector after passing through an anti-scatter grid aiding in noise reduction. The radiographic appearance of the breast varies depending on the relative distribution of adipose and glandular tissue, with the first being radiologically lucent and appearing dark and the latter being characterized as radiologically dense and appearing brighter in the mammographic image. These variations are associated with breast cancer risk (Boyd et al 1998, Li et al 2010).
In Fig. 3.2 examples of different breast types with varying densities are illustrated.

The last years conventional imaging is replaced by digital imaging the same happens in mammography. Screen-film mammography (SFM) was the standard technique for breast cancer screening and diagnosing for many years, but nowadays a two view (medio-lateral oblique and cranio-caudal) examination using full-field digital mammography (FFDM) is preferred. One of the advantages of digital imaging is the implementation of advanced applications, including computer-aided diagnosis (CAD) (Costaridou et al 2008, Karahaliou et al 2012).

Doctors use a standard system to describe mammogram findings and results. This system (called Breast Imaging Reporting and Data System, BI-RADS) sorts the results into categories numbered 0 (incomplete) to 6 (known biopsy - proven malignancy), in order to standardize reporting, reduce confusion in breast imaging interpretations, and facilitate outcome monitoring (ACR 1998, Birads 2003). In Table 3.1 the classification of BIRADS is presented.

<table>
<thead>
<tr>
<th>BI-RADS Category</th>
<th>Assessment</th>
<th>Clinical Management Recommendation(s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>Assessment incomplete</td>
<td>Need to review prior studies and/or complete additional imaging</td>
</tr>
<tr>
<td>1</td>
<td>Negative</td>
<td>Continue routine screening</td>
</tr>
<tr>
<td>2</td>
<td>Benign finding</td>
<td>Continue routine screening</td>
</tr>
<tr>
<td>3</td>
<td>Probably benign finding</td>
<td>Short-term follow-up mammogram</td>
</tr>
<tr>
<td>4</td>
<td>Suspicious abnormality</td>
<td>Biopsy needs to be performed</td>
</tr>
<tr>
<td>5</td>
<td>Highly suggestive of malignancy</td>
<td>Biopsy and treatment are necessary</td>
</tr>
<tr>
<td>6</td>
<td>Known biopsy-proven malignancy</td>
<td>Assure and follow-up the response to treatment</td>
</tr>
</tbody>
</table>

3.3. Breast abnormalities

X-ray imaging of benign or malignant breast pathologies results in mammographic findings usually in the form of masses and microcalcifications (μCs) (Yaffe 2003). These abnormalities are described according to the BIRAD lexicon as follows:
Mass

A “Mass” is a space occupying lesion that can be identified in two different projections. If a potential mass is seen in only a single projection it should be called a “density” until its three-dimensionality is confirmed. Masses can be categorized according to the shape of their margins as follows:

![Fig. 3.3: a) Circumscribed mass and b) mass with spiculated margins](image)

Circumscribed (well-defined or sharply-defined) margins: This mass is characterized by well defined contour between the lesion and the surrounding breast tissue. Usually this contour has an ellipsoid shape as shown in Fig. 3.3a.

Indistinct (ill defined) margins: The poor definition of the margins raises concern that there may be infiltration by the lesion and this is not likely due to superimposed normal breast tissue.

Spiculated margins: The lesion is characterized by lines radiating from the margins of a mass (Fig. 3.3b).

Calcifications

The diagnostic approach to breast calcifications is to analyze the morphology, distribution and sometimes change over time. The form or morphology of calcifications is the most important factor in deciding whether they are typically benign or not. If not, they are either suspicious or of high probability of malignancy. In this case biopsy is usually needed to determine the etiology of these calcifications.

Benign calcification: Many calcifications can be classified as typically benign and in this case there is no need for follow up (i.e. BIRADS 1 or 2) (Fig. 3.4). An example of such benign lesions are skin calcifications. These are usually lucent-centered details. Atypical forms may be confirmed as skin
calcifications by tangential views. Usually they are located along the inframammary fold parasternally and in the axilla and areola. Another example is the vascular calcifications that can be identified through their arrangement in lines.

![Fig. 3.4: Benign calcifications a) skin calcification b) coarse heterogeneous calcification](image)

Suspicious calcifications: If calcifications are not typically benign, they are either 'Suspicious or of Intermediate Concern' or 'High Probability of Malignancy'. Suspicious calcifications have either an amorphous or coarse heterogeneous form (Fig. 3.5a and b). Usually for these calcifications a biopsy is needed to determine their exact nature.

Amorphous or indistinct calcifications: they are defined as 'without a clearly defined shape or form'. These calcifications are usually so small or hazy in appearance, that a more specific morphologic classification cannot be determined.

Coarse heterogeneous calcifications: formerly called coarse granular, these are irregular, conspicuous calcifications that are generally larger than 0.5 mm. They are considered to be of intermediate concern, along with amorphous microcalcifications.

High probability of malignancy: Calcifications with a higher probability of malignancy are fine pleomorphic and fine linear or fine linear branching. Two images of those two types of high probability of malignancy are shown in Fig. 3.5c and d.
Fine pleomorphic: These calcifications vary in size and shape and are usually more conspicuous than the amorphic calcifications. There is a 25-40% risk of malignancy.

Fine linear or fine linear branching (casting) calcifications: These are thin, linear or curvilinear irregular calcifications. They may be discontinuous with an appearance suggesting filling of the lumen of a duct, i.e. 'casting' calcifications. These calcifications are classified as BIRADS 5.

3.4. Breast Tomomosynthesis

3.4.1. Principles of tomosynthesis
Breast tomosynthesis (BT) is a 3D imaging technology that involves acquiring low-dose images of a stationary compressed breast at multiple angles during a short scan. These individual images are then reconstructed into a series of high-resolution slices parallel to the detector plane that can be displayed individually (Smith 2008). The problems caused by dense breast tissue, overlapping structures, and thus structure noise in a single slice 2D mammography imaging are significantly reduced in reconstructed tomosynthesis slices. The switch from screen-film to electronic means of image acquisition simplified
significantly the tomosynthesis process; however, it could not compete with Computed Tomography (CT) which had become widely accepted by the late 70s. The interest in BT was regained in the late 90s with the appearance of digital flat-panel detectors that could result in high quality images at rapid readout rates (Dobbins and Godfrey 2003).

In BT, the x-ray source is rotated over a limited arc angle while the breast is compressed in a standard way. A series of low dose exposures are acquired, creating a series of digital images. The images are actually projections of the breast at different angles which are reconstructed into thin slices (Smith 2008). Slices may be as thin as 0.5 mm, but are usually 1 mm. The reconstructed BT slices provide a pseudo-3D structural information and may reduce the masking effects of fibroglandular tissues (Park et al 2007). Tomosynthesis can prove to be very useful for imaging the breast in an attempt to optimize the detection and depth localization of abnormalities with dose levels not higher than those of conventional mammography (Niklason et al 1997). Recent studies have shown that BT can be beneficial in the characterization of lesions particularly in dense breasts (Gong et al 2006) and has the potential to reduce the recall rate (Gur et al 2009) and 3D lesion localization (Smith 2008).

X-ray spectrum is an important aspect in dose optimization. The spectrum reported in BT is usually the same as the one used in conventional digital mammography, but the optimum spectrum depends on whether BT will be used for the detectability of both masses and calcifications or just for masses (Dobbins 2009). Typically, W target is used along with Al, Ag or Rh filters and for an average 50% glandularity and thicknesses between 2 to 6 cm, tube voltage varies from 26 to 33 kVp. The choice of the harder energy spectrum with target/filter combination of W/Rh, with Rh thickness of 50μm, has reported higher signal to noise ratio over Mo spectrum and seems to be more beneficial for tomosynthesis (Zhao et al 2005).

In general, doses in a single view BT exam are about one to two times that of a single view FFDM (Dobbins 2009). Specifically, the mean glandular doses (MGD) for BT acquisition are reported to vary between 0.6 to 4.0 mGy (Feng and Sechopoulos 2012, Poplack et al 2007, Dobbins 2009).

In most studies, the angular range varies from 30°- 60°, whereas the number of projection images is usually between 11-25 (Sechopoulos and Ghetti 2009, Suryanarayanan et al 2000, Wu et al 2003, Poplack et al 2007). Several manufacturers have applied different methods to develop and perform tomosynthesis. Manufacturers vary the arc of movement, the number of individual exposures, use of continuous or pulsed exposure, stability or movement of the detector, exposure parameters, total dose, effective size of pixels, x-ray source/filter source, single or binned pixels, and patient position (Helvie 2010).
There are two main types of scan motion: ‘step and shoot’ and continuous motion. ‘Step and shoot’ scans may reduce blur due to tube motion during the acquisition but have higher processing time to perform the whole scan and there may also be additional problems arising from vibration of the tube as it stops and starts. There is however little published research on the impact of changing the type of scan motion on BT images.

The projection images acquired on the detector are then reconstructed in order to obtain the three-dimensional visualization of breast. Image interpretation and display provide additional challenges. The principle is to perform appropriate movement of the source and the detector during the acquisition and suitable shifting and transformations in the projected images, in order to keep the plane of interest being projected on the same place on the detector. The result is that the plane of interest remains in focus while all the rest out of focus details appear blurred on the obtained tomogram. Image reconstruction is usually performed by filtered back projection. However, several reconstruction techniques have been developed, including the Back Projection, Multiple Projection Algorithm (Kolitsi et al 1992), iterative methods and other advanced reconstruction approaches like Algebraic Reconstruction Technique, Matrix Inversion tomosynthesis and Maximum Likelihood method (Suryanarayanan et al 2000, Wu et al 2003).

Similarly, several filtering methods have been proposed to be used with BT. These filters are applied on the projection images either prior to the reconstruction or afterwards on the reconstructed planes. Noise in BT comes from a variety of sources some of which can be neglected (electronic noise) with proper system design and operating conditions, while some others cannot be eliminated, such as the photon noise. Another type of intrinsic noise, is the X-ray quantum noise coming from the statistical variation of the x-ray photon flux which is Poisson distributed. This noise cannot be avoided, only slowly reduced with increasing photons (Adachi et al 2002, Buliev 2004). Tomograms also suffer from noise due to blurred out of focus anatomical structures that exist in the tomograms or from artifacts and distortion of the details’ shape due to the limited angle nature of BT and insufficient number of projections that allows for a recovery of a limited space in the Fourier domain. Paying careful attention to the choice of acquisition parameters and the use of appropriate deblurring algorithms can minimize these types of artifacts (Dobbins and Godfrey 2003). Spatial frequency filtering has been investigated as a noise removal method. One approach involves direct application of high pass filters to the original projection images prior to the reconstruction (Edholm and Quiding 1970). Thus the images are cleared from low-frequency blurry artifacts and superimposed tissue. A well-known high-pass filter is the ramp filter widely used in tomosynthesis.
3.4.2. Systems and acquisition technologies

There is a variety of motion geometries for BT systems that manufacturers apply into their equipments.

![Diagram of tomosynthesis geometries using isocentric motion](image)

**Fig. 3.6: Tomosynthesis geometries using isocentric motion.** (a) Complete isocentric motion (b) Partial isocentric motion and (c) Partial isocentric motion of (Niklason et al 1997) in which the detector is stationary. (Figure adapted from figure 5 in (Dobbins and Godfrey 2003)).

Most tomosynthesis geometries of motion involve having the x-ray tube move in a path parallel to the plane of the image receptor. However, there are certain geometries adopted in tomographic devices, like the C-arm device, where only the source or both the source and the detector move in an arc about a point called the center of rotation or isocenter. The two basic motion geometries for BT systems can be divided in two categories: complete isocentric motion (Fig. 3.6a), where the x-ray source and the detector are fixed with respect to each other and rotate about the same axis, and partial isocentric motion. In partial isocentric motion the x-ray source rotates in a small arc whereas the detector either moves horizontally (Fig. 3.6b) or remains stationary (Fig. 3.6c). The partial isocentric motion with stationary detector is the most typical tomosynthesis geometry for mammographic applications due to fact that the mechanical construction of the current devices used in conventional mammography can easily incorporate this type of motion.

Currently, most of the BT systems have the same basic components as the digital mammography systems in terms of type of detector, breast support, compression plates, x-ray tube mounted on an arm, making these systems suitable for working in both modalities with small adjustments. The most important adjustments include the ability of the x-ray tube to rotate around a specific point between the source and the detector and the use of different x-ray spectrum filtration suitable and optimized for BT. The characteristics and specifications of the BT systems in use or prototypes under development, including
those diverging from the typical design described above, are summarized in a recent review by Sechopoulos (Sechopoulos 2013). In Table 3.2, the specifications of the two BT commercial systems that are currently in use, but still not FDA approved for clinical use in the U.S., are presented.

Table 3.2: Characteristics of BT systems in clinical use

<table>
<thead>
<tr>
<th></th>
<th>Siemens MAMMOMAT Inspiration</th>
<th>Hologic Selenia Dimensions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acquisition arc</td>
<td>50°</td>
<td>15°</td>
</tr>
<tr>
<td># of projections</td>
<td>25</td>
<td>15</td>
</tr>
<tr>
<td>X-ray tube</td>
<td>W, 0.05 mm Rh</td>
<td>W, 0.7 mm Al</td>
</tr>
<tr>
<td></td>
<td>continuous motion</td>
<td>continuous motion</td>
</tr>
<tr>
<td>Detector</td>
<td>Full field direct (a-Se)</td>
<td>Full field direct (a-Se)</td>
</tr>
<tr>
<td></td>
<td>size (cm): 24x30</td>
<td>size (cm): 24x29</td>
</tr>
<tr>
<td></td>
<td>pixel size (μm): 85</td>
<td>pixel size (μm): 70</td>
</tr>
<tr>
<td></td>
<td>stationary</td>
<td>(bined 2x2)</td>
</tr>
<tr>
<td>SOD (cm)</td>
<td>65.6</td>
<td>70</td>
</tr>
<tr>
<td>IDD (cm)</td>
<td>4.7</td>
<td>0</td>
</tr>
<tr>
<td>Air gap (cm)</td>
<td>1.7</td>
<td>2.5</td>
</tr>
<tr>
<td>Reconstruction</td>
<td>FBP</td>
<td>FBP</td>
</tr>
</tbody>
</table>

\(a\) (Sechopoulos 2013, Baker and Lo 2011, Shaheen et al 2011, Young et al 2009)

\(b\) (Sechopoulos 2013, Baker and Lo 2011)

\(c\) Reconstruction results in an in-plane pixel size of approximately 100μm

3.4.3. Reconstruction algorithms

Shift and Add algorithm (SAA)

The Shift and add (SAA) method is a fundamental reconstruction algorithm that has been modified by many researchers for tomosynthesis application. An example is the application of this algorithm to mammography, to line up each projection image after stretching it along the direction of x-ray tube’s motion and shifting it by an appropriate amount (Niklason et al 1997). As a method, it reconstructs planes that are parallel to the plane about which the source and detector moves, called fulcrum plane (Maravilla et al 1983, Haaker et al 1985b, Rimkus et al 1989). According to this method structures from planes below or above the fulcrum plane can be reconstructed by shifting the acquired projection image prior to adding. As a result, structures from this specific plane come in focus and appear sharp in the final tomogram, whereas structures from all the other planes with different positions in the shifted images become blurred after adding. The SAA algorithm also serves as a solid foundation for further deblurring algorithms in breast tomosynthesis field (Chen et al 2004).
**Back Projection algorithm (BP)**

The backprojection algorithm is a direct algorithm that can be applied in tomosynthesis among other modalities to reconstruct arbitrary planes of any direction in relation to the fulcrum plane (Kampp 1986). The theorem of projections, also called a center cut theorem (Central Slice Theorem-CST), facilitates the process. According to CST, a 1D Fourier transform of the Radon transform of the image corresponds to the 2D Fourier transform of that image, where each pixel in the projection images is transferred backwards and deposited at the intersection of the x-ray source with the reconstructed plane. The BP can reconstruct planes of any desired direction; however, since it is performed pixel-by-pixel, it results in long reconstruction times. A subcategory of this algorithm is the filtered backprojection (FBP), which works with the same way but a filter (usually ramp filter) is applied on the projections prior to reconstruction. The FBP is the most widespread algorithm for CT reconstructions and the one that most BT systems have in clinical use nowadays. Finally, it has low processing time compared with other methods.

**Multiple Projection algorithm (MPA)**

The Multiple Projection algorithm is a BT reconstruction algorithm of maximized efficiency capable of producing images of arbitrary orientations and is described in detail in Kolitsi *et al.* (Kolitsi *et al* 1992). In the case of MPA, the inverse problem facilitates groups of pixels (rows) instead of one single pixel, thus resulting in a multifold reduction of processing time. For each acquired angle, the projection data is first geometrically projected onto the “horizontal plane”, a plane parallel to the reconstruction plane, passing through the intersection of the beam’s central axis with the detector plane. Then, this projection image is shifted by an amount depending on the position of reconstructed plane, and normalized to the magnification of the plane of the isocenter. All these images are added up to depict the plane of interest with a modified size. This image is formed on a plane parallel to the reconstruction plane, called “image formation plane”. MPA can reproduce planes tilted at any desired angle with respect to x-axis with an important reduction in the reconstruction time. However, tilt angles with respect to y-axis are limited to less than 45°.
Iterative reconstruction techniques

Iterative reconstruction algorithms were primarily introduced in nuclear medicine (SPECT, PET) and have been recently introduced in multidetector CT scanners to reduce noise. These algorithms differ from direct one like FBP since the reconstruction image is not given by a single reconstruction step but through cycles of calculations. Actually, they are methods based on an initial estimation for the imaging object that is being iteratively corrected in order to match the real object. Iterative algorithms allow obtaining a better reconstruction compared with direct methods at the cost of higher computational time. There are a lot of iterative reconstruction techniques, the most widely used being: algebraic reconstruction techniques (ART), the simultaneous iterative reconstruction technique (SIRT), the simultaneous algebraic reconstruction technique (SART) and the iterative least squares technique (ILST) (Colsher 1977).

3.4.4. Evaluation metrics

For the quantitative evaluation of the images acquired in mammography and the reconstructed features in BT, several evaluation metrics have been introduced through time by researchers. In this chapter, the evaluation metrics that were used for the needs of this study will be presented. The most commonly used is the Contrast to Noise ratio (CNR) (Wagner 1977, Zhang et al 2006, Bliznakova et al 2010, Lu et al 2011, Malliori et al 2012) defined as:

\[
CNR = \frac{|I_{obj} - I_{back}|}{\sigma_{back}} \quad \text{Equation 3.1}
\]

where \(I_{obj}\) is the average value of the object under investigation, whereas \(I_{back}\) and \(\sigma_{back}\) are the average value and standard deviation of the background, respectively. There are several variations of CNR reported in studies, using in the denominator the standard deviation of both the background and the feature, (Shaheen et al 2011, Van De Sompel et al 2011) or the product of mean value and standard deviation of the background (Sechopoulos and Ghetti 2009).

The CNR as it was presented is consistent with the definition of Rose’s ‘signal to noise ratio’ (Rose 1948) measuring the detectability of a feature in a reconstruction plane and is often reported in studies as signal difference to noise ratio (SDNR or SNR) (Wu et al 2004, Samei et al 2005, Zhou et al 2007, Mertelmeier et al 2008, Sarkar et al 2009).
where \(I_{obj}, I_{back}\) and \(\sigma_{back}\) are the same values as in the CNR and \(\sigma_{obj}\) is the standard deviation of the object. For the case of phase contrast imaging and the detection of features with edge enhancement, the equation of SNR can be written as:

\[
SNR_{\text{edge}} = \frac{I_{\text{max}} - I_{\text{min}}}{\sqrt{2 \cdot \sigma_{\text{back}}}} \quad \text{Equation 3.3}
\]

This metric takes into consideration the maximum \(I_{\text{max}}\) and minimum \(I_{\text{min}}\) values appearing in the edges of an object as a result of phase contract effect and not the mean values, whereas \(\sigma_{\text{back}}\) is again the standard deviation of the background.

In this study the objects under investigation are usually masses and \(\mu\)Cs. As a result, regions of interest inside each of those objects are defined for the calculation of each evaluation metric. For phase contrast imaging fibers are also under consideration and the ROI in this case is a square that covers both edges of the object. Also the background ROIs are chosen to be close to the object under investigation covering a larger area with no mammographic details into it.

Another metric that is used is the Contrast (C) (Wu et al 2003, McKinley et al 2004, Samei et al 2005, Shaheen et al 2011) defined as:

\[
\text{Contrast} = \frac{I_{obj} - I_{back}}{I_{back}} \quad \text{Equation 3.4}
\]

Contrast may be also given as a percentage (%). Contrast does not make use of standard deviations; thus there is no estimation of noise in this metric. In digital mammography, CNR is a more relevant parameter for image quality compared to C (Samei et al 2005, Borg et al 2012, Van Engen et al 2006). As for CNR, C has also a different version, where in the denominator the sum of the mean values of both the background and the object, could be used (Michelson 1927).

For BT imaging, where objects appear in focus at different planes, the Artifact Spread Function (ASF) is a valuable metric for in depth quality evaluation. Moreover, the ASF quantifies the “out-of-plane” artifacts observed in planes outside the in focus plane (Zhang et al 2006, Lu et al 2011):

\[
ASF(z) = \frac{CNR(z)}{CNR(z_0)} \quad \text{Equation 3.5}
\]
where $z_0$ is the location of the in focus plane and $z$ is the location of the off focus plane, respectively. A quality factor combining in one metric the in-plane image quality and the vertical resolution and defined as the CNR over the ASF was proposed by Sechopoulos, (Sechopoulos and Ghetti 2009).

Finally, in order to compare images acquired from different experiments in terms of equal doses, the following figure of merit (FOM) has been used (Arfelli 2000):

$$FOM = \frac{\text{CNR}}{\sqrt{\text{MGD}}} \quad \text{Equation 3.6}$$

This FOM uses the CNR and the mean glandular dose (MGD) and is based on the fact that for a given beam quality, if the system is quantum limited, CNR is proportional to the square root of the incident exposure to the breast (Samei et al 2005), based on Poisson distribution and statistics or to the mean glandular dose (MGD) (Borg et al 2012). In some studies, the entrance surface exposure (ESE) is used instead of the MGD (McKinley et al 2004, Samei et al 2005).

4. **BREAST AUGMENTATION**

Breast augmentation, also known as augmentation mammoplasty, is a plastic surgery term for the breast implant mammoplasty approaches used to increase the size, change the shape and alter the texture of woman's breast. Breast augmentation is one of the most frequently performed procedures in plastic surgery. Approximately 400,000 women choose to undergo breast augmentation in the United States each year, making it the second most popular cosmetic procedure in the United States after liposuction.

Since the late nineteenth century, breast implants have been used to surgically augment the size (volume), modify the shape (contour), and enhance the feel of woman's breasts. Vincenz Czerny, a surgical pioneer in breast reconstruction, performed the earliest breast implant emplacement in 1895, when he used the patient's autologous adipose tissue to repair asymmetry of the breast from which he had removed a tumor (Czerny 1895). Furthermore, throughout the 1950s and the 1960s, plastic surgeons used synthetic fillers including silicone injections received by 50,000 women, from which some developed silicone granulomas and breast hardening that required treatment by mastectomy (Anderson 1997).

Breast reconstruction is a type of surgery for women who have had all or part of a breast removed. The surgery rebuilds the breast mound to match the size and shape of the other breast. The nipple and the darker area around the nipple (areola) can also be added. Most women who have had a breast removed
(mastectomy) are able to have breast reconstruction. Women who have had only the part of the breast around the cancer removed (lumpectomy or breast-conserving surgery) may sometimes need reconstruction.

4.1. Breast implants

Nowadays, there are several types of breast implants that doctors use for breast reconstruction and augmentation of a woman's breast. They can be divided in three types, defined by the filler material: saline, silicone, and composite. Saline breast implants have an elastomer silicone shell filled with sterile saline solution. Silicone breast implants have an elastomer silicone shell filled with viscous silicone gel. Some newer types use thicker silicone gel, called cohesive gel. The thickest ones are sometimes called “gummy bear” implants and are made of highly cohesive silicone. They are more accurately called form-stable implants, meaning that they keep their shape even if the cover is cut or broken. Although it was first thought they would not leak even if they did break, there have been reports of ruptures with leakage. Both silicone and saline implants are FDA approved for breast augmentation. Alternative composition breast implants featured miscellaneous fillers, such as soy oil and polypropylene string. Alternative breast implants that have different shells and are filled with different materials are being studied, but one can only get them when they are available in clinical trials.

Complications

The plastic surgical emplacement of breast-implant devices, either for breast reconstruction or for aesthetic purpose, presents the same health risks common to surgery, such as adverse reaction to anesthesia, hematoma (post-operative bleeding) and seroma (fluid accumulation). Complications specific to breast augmentation include breast pain, altered sensation, impeded breast-feeding function, visible wrinkling, asymmetry, thinning of the breast tissue, and symmastia, the “bread loafing” of the bust that interrupts the natural plane between the breasts. Two of the most common complications of breast implant are the implant rupture and the capsular contracture. The suspected reasons of breast implant rupture are: damage during surgical procedures, chemical degradation of breast implant shell, trauma and mechanical pressure during mammography procedure. Capsular contracture occurs when the collagen-fiber capsule thickens and compresses the breast implant. It is a painful complication whose cause is unknown and might distort the implant or the breast or both. Specific treatments for the complications of capsular contracture and capsular rupture are periodic MRI monitoring and physical examinations.
In addition, there are other drawbacks of breast implants especially in breast imaging that this thesis investigates and are going to be discussed in next chapters. Apart from the obscuring glandular tissue, breast implants may include changes that interfere with early cancer detection. Compression of breast tissue by the implant makes it more difficult to identify some of the early changes in architectural pattern that may be seen with cancer. Calcifications in capsular tissues may cause false-positive findings (Pennisi et al 1977). Diverticula of the capsule can present as "palpable breast masses" following augmentation. Rupture of the implant with extrusion of the silicone gel into the glandular tissue may present as a nodule. Finally, a study suggested that augmented patients present with more advanced disease when breast cancer is diagnosed and have a higher percentage of invasive lesions and positive axillary nodes, thus having a poor prognosis (Silverstein et al).

4.2. Incision and implant placement

There are multiple options regarding the surgical access for placing the breast implant. Selection of an access is primarily based on the implant size, type and location. Furthermore, skin quality and residual breast tissue also influence the decision for the surgical access to be used. Prior breast surgery or breast deformities also limit the possible options. The typical incision regions for breast augmentation are:

- along the underside of breast (infra mammary)
- under the arm (transaxillary)
- around the nipple (periareolar)
- through the mastectomy scar (for reconstruction)

The FDA-approved labeling warns surgeons not to place breast implants through the belly-button (periumbilical approach). The location of the incision can affect how visible the scars are, as well as any complications after surgery.

Placing the implant through an incision under the arm will likely require the surgeon to use an endoscope, a tool with a camera and other surgical instruments inserted into the incision site to help the surgeon guide the implant into place. While there will likely be no visible scar around the breast, the scar on the underside of arm may be visible. In addition, cutting around the edge of the nipple (areola) may cause problems with loss or change of sensation in the nipple.

In the case of silicone gel filled implants, they are already filled with silicone gel when inserted, whereas in the case of not pre-filled saline filled implants the surgeon inserts the silicone shell and then fills the implant to the desired level with saline. The incision is then closed with stitches. The surgeon may place temporary drains in the incision prior to closing it to prevent fluid or blood accumulation. Catheters to
deliver pain medicine at the site of the incision may also be placed prior to closing the incision. The drains or catheters would be removed during a follow-up visit after surgery.

Breast implants may be placed beneath the tissue of the breast (subglandular), or even deeper, beneath the muscle of the chest wall (totally submuscular). There is also a subpectoral (or partially submuscular) placement that some surgeons use, meaning that the implant is placed beneath the pectoralis major muscle only, leaving the outer, lower 1/3 of the implant not covered by muscle tissue. The decision about the placement is determined by the natural shape of the woman's breasts and chest wall muscles.

Fig. 4.1: Three main positions for breast implant placement.

Whenever possible, surgeons prefer to place the implants in a totally submuscular position since the advantages of placing the implants completely beneath the muscle of the chest wall greatly improves the long-term appearance of the breasts following augmentation. This is for three main reasons. First, mammography is easier and the quality is better when the breast implant is separated from the breast tissue by the layer of muscle. Secondly, there is a lower incidence of capsular contracture when the implants are placed totally beneath the muscle. Capsular contracture occurs when the body produces scar tissue around the implant. This may change the shape of the implant and make the breasts asymmetric. Finally, placing the implants beneath the muscle of the chest wall makes the breasts appear more natural because there is more breast tissue covering the implant.

4.3. Breast Implant and Mammography

Several characteristics of silicone gel implants and the techniques of their placement affect the x-ray based imaging of the breast. The presence of silicone gel-filled implants interferes with standard
mammography, since silicone is a radiopaque material. The physical presence of the implant compresses fat and glandular tissues increasing the density of the breasts, which frequently lacks the contrast needed to detect subtle early features associated with breast cancer (Eklund et al 1988). This means that in the case of implant insertion more tissue may be imaged in a smaller space, which causes superposition of structures, resulting in poorer image quality. Since silicone filled implants have a low x-ray transmission in the region of the implant, the detection of small masses in the breast is reduced (Brody 1989, Martin 1989). As a result, breast lesions near the silicone gel implant require special mammography procedures for their detection (Arthur et al 1991). An ‘Eklund’ view mammogram is often performed to enhance breast tissue visualization (Fig. 4.2b on the right). When using the Eklund technique the breast implant is pushed against the chest wall and the breast tissue is distracted forward, away from the implant, to improve the image of the breast tissue. Despite the use of displacement techniques, as much as one-third of a woman’s breast tissue may not be adequately visualized by mammography when breast implants are present. Subpectoral implants, compared with subglandular implants, may offer improved mammographic imaging of the breast tissue because the muscle functions as a barrier that partially separates the implant from the breast tissue, thereby minimizing overlap interference. The ratio of the amount of breast tissue to implant volume also plays a role in how easily the breast tissue is visualized by mammography. Breast tissue imaging will be reduced when there is a small amount of breast tissue and larger implants because it will be more difficult to separate the breast tissue away from the implant using displacement techniques.

Moreover, silicone gel implants obscure portions of the breast on mammography images (Cohen et al 1997, Gumucio et al 1988). It has been estimated that about 25% of the breast tissue is not visible in the presence of silicone gel implants on mammography images (Wolfe 1978). Studies taking into account the
visualized mammography breast tissue area before and after augmentation mammoplasty indicate a reduction of measurable tissue area in the range of 15%–44% depending on the imaging procedure and the positioning (Silverstein et al 1990). Anterior breast tissue was visualized better with displacement mammography, while compression mammography indicated better results for the posterior breast tissue.

Breast implants tend to be “radiographically opaque” on x-ray film: they appear whiter because they block some of the passage of the x-ray beam. Silicone gel implants appear whiter than saline implants because they block more of the x-ray beam. Women with breast implants require special mammographic techniques to image their breast tissue. The presence of breast implants may make mammography more difficult, and the breast implant may obscure or limit the visualization of breast tissue or the detection of breast cancer. Saline breast implant rupture can occur from breast compression during mammography, although this is rare. Displacement techniques reduce compression of the implant, and the subsequent risk of capsule or device rupture. Saline implant failure during mammography is less likely to occur with smooth implants that are under the muscle, and in the absence of capsular contracture. In some situations where conventional mammography is not able to visualize breasts, specialized mammography, ultrasound and/or MRI studies may be necessary to evaluate breast lumps or the condition of the implants.

4.4. Physical Properties of Silicone Gel

This work focuses on silicone implants, due to their wide acceptance from the medical community and the problems they introduce to diagnostic imaging (Dutteille et al 2014, American Society for Aesthetic Plastic Surgery 2012, ISAPS). Specifically, silicone implants are filled with viscous silicone gel and covered with silicone polymer. Silicone gel, used for breast implants, is a synthetic material inert containing 38% silicon (Si) usually in the form of a silicone tetramer (polydimethylsiloxane, PDMS) with chemical composition: $CH_3[Si(CH_3)2O]_4Si(CH_3)_3$ that has an effective atomic number of 10.37, a density of 0.97 g/cc and a volume ranging between 90 cc and 800 cc. In Fig. 4.3 some of the physical quantities of silicone gel are illustrated in comparison with water, such as the attenuation and absorption coefficients (coherent scattering, photoelectric absorption and total attenuation coefficient). Water is a good mimicking material for breast tissue with an averaged density and in a lot of studies is used as a substance material for breast imaging investigations.
The squares in Fig. 4.3 illustrate the range of energies (20 keV-30 keV) applied in conventional mammography. A comparison of this two squares shows that silicone gel is a material with higher absorption and attenuation values meaning that has low x-ray transmission. This difference in their values is increasing with the decrease in energy whereas it starts to fade at higher energies. As a result silicone is a radiodense material characterized by low x-ray transmission which can interfere with early detection of breast lesions.

Fig. 4.3: Diagrams of the coherent scatter, photoelectric absorption and total attenuation for water on the left and silicone gel on the right.

Fig. 4.4: Attenuation coefficient diagram for breast tissue, breast lesions and silicone gel
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Part 2

Experimental Work

This part is dedicated to the simulated and experimental studies of this thesis. The imaging drawbacks of the silicone gel insertion into breasts in Mammography and Breast Tomosynthesis are examined. A wide range of software and hardware phantoms are introduced for image quality evaluations. The detectability of high and low contrast lesions in the presence of silicone gel of different thicknesses is investigated. Alternative imaging modalities are studied as possible techniques for improved image quality of augmented breasts. Dual Energy, Subtraction Mammography and Tomosynthesis, Breast Tomosynthesis in a discontinuous arc, as well as an extended study of phase contrast imaging are assessed.
5. EVALUATION OF THE EFFECT OF SILICONE BREAST INSERTS ON MAMMOGRAPHY AND BREAST TOMOSYNTHESIS IMAGES: A MONTE CARLO SIMULATION STUDY

5.1. Introduction

Breast cancer screening and early diagnosis is the primary aim of the x-ray mammography and breast tomosynthesis (BT). On the other hand, breast augmentation is one of the most popular cosmetic surgeries worldwide. Patients with silicone gel implants should be treated with an alternative way regarding the dose limits, the incident beam energy and the acquisition protocols. However, prostheses used for breast reconstruction and augmentation contain a material with higher atomic number ($Z$) compared to the breast tissue and this may affect the image quality and the dose distribution within the breast.

There are several types of breast implants used for breast augmentation and reconstruction in plastic surgery as they are described in Chapter 4.1. This work focuses on silicone implants, due to their wide acceptance from the medical community and the problems they introduce to diagnostic imaging (Duteille et al 2014, American Society for Aesthetic Plastic surgery 2012, ISAPS). Specifically, in the simulations of this study a silicone tetramer (polydimethylsiloxane) with chemical composition: $\text{CH}_3[\text{Si}($$\text{CH}_3$)$_2\text{O}]_4\text{Si}($$\text{CH}_3$)$_3$ that has an effective atomic number of 10.37 and a density of 0.97 g/cc.

The way in which the implant presence affects diagnostic breast imaging in combination with its high percent of usage make further investigation on this issue necessary. In Chapter 4.3 an extended summary of the complication in breast imaging in the case of augmented breasts is described (Brody 1989, Martin 1989, Cohen 1997, Gumucio et al 1988). The objective of this work is to further investigate the effect of the silicone gel insertion on the image quality of mammography and BT images considering different energies and dose levels. Moreover, this study evaluates the detectability of both high and low contrast features on simulated images in the presence of silicone gel implant of different thickness at beam energies in the interval of 20-30 keV. For his purpose, three software phantoms with silicone gel prosthesis were designed and used for x-ray imaging simulations. X-ray transport and images of these phantoms were generated in a mammographic and BT mode, using the validated in-house developed software XRayImagingSimulator (Bliznakova et al 2010, Lazos et al 2000). X-ray mammography and BT imaging processes were modeled and simulated with incident energies as well as dose limits modified in a way to implement constant incident photon fluence for all the simulations. Six monochromatic x-ray beams with energies in the interval 20-30 keV were simulated resulting in entrance surface exposure
(ESE) in the range of 1.83 to 4.32 mGy for the first and third phantom and 3.59 to 8.47 mGy for the second phantom. The visibility of breast masses and the detection of μCs under different silicone base thickness were evaluated versus the incident beam energy and the type of the x-ray imaging technique (mammography and BT).

5.2. Software phantoms

Three different software phantoms dedicated for silicone gel implant breast imaging were designed using the phantom module of the XRayImagingSimulator (Bliznakova et al 2010). The first one has a step-wedge geometry, the second one has a helical form, while the third is a breast-like phantom. The first two phantoms were designed with escalating geometries in order to investigate the effects of different silicone thicknesses in a single imaging phantom. The first one has silicone inserts reaching a thickness of 36 mm, while the second one gives the ability to investigate higher thicknesses of silicone up to 49 mm in more detail. In addition, the third phantom was designed to yield a more realistic illustration of breast implant insertion. Breast abnormalities, i.e. microcalcifications (CaCO$_3$) and low contrast features (breast masses) with nominal densities of 2.8 g/cm$^3$ and 1.011 g/cm$^3$, respectively, were simulated and inserted near the silicone gel based implant material. These phantoms were the basic tools in investigating the effect of silicone gel implants on the image quality and breast lesion detection under different imaging conditions.

**Step-wedge phantom**

A phantom, shown in Fig. 5.1(a), was designed using 18 adjacent cuboids and modeled from a silicone gel (CH$_3$[Si(CH$_3$)$_4$O]$_2$Si(CH$_3$)$_3$), with a thickness in the range of 2 mm-36 mm, forming a step-wedge (geometry). The step-wedge phantom was placed in a 50×100×40 mm$^3$ homogeneous block, filled with a mixture of 50% adipose and 50% glandular tissue with a density of 0.982 g/cm$^3$ (Poletti et al 2001). A CaCO$_3$ sphere with a radius of 0.2 mm was placed at a distance of 2mm under each cuboid. The phantom was converted to a voxel-based one, whereas – for time consuming reasons – a voxel size of 200 µm was chosen along each direction.

**Snail phantom**

A snail phantom, shown in Fig. 5.1(b), was created consisting of 49 adjacent 2×2 mm$^2$ silicone gel cuboids (representing the implant) with thicknesses ranging from 1 mm to 49 mm. In addition, 49 spheres simulated as CaCO$_3$ with a radius of 0.2 mm, were placed at a distance of 1 mm under each implant cuboid. Modeled lesions and implant were inserted within a homogeneous block with size of 18×18×54
mm³ simulated as a mixture of 50% adipose and 50% glandular tissue. In this phantom the thickest implant material is placed at the center of the phantom in contrast to the previous step-wedge one, where the thickest slab is placed at the right side of the phantom. Similarly to the previous phantom, the snail phantom was converted to a voxel-based one with a voxel size of 100 μm along each direction.

**Phantom with a realistic implant shape**

A rectangular slab phantom with a realistic shape and composition mimicking a silicone gel implant was designed as shown in Fig. 5.1(c). The phantom with dimensions of 50x100x45 mm³ was composed of a homogeneous mixture of material simulating 50% adipose tissue and 50% glandular tissue. The implant was modeled in the form of a semi-ellipsoid with dimensions of 20x46x20 mm³ filled with silicone gel. Two clusters of μCs, each one consisting of six CaCO₃ spheres with radii of 0.138 mm and 0.4 mm respectively, and a set of three water spheres with radii of 1mm, 1.5mm and 2mm, respectively, were inserted within the base material. The construction of this phantom allows investigations of both high (μCs) and low (masses) contrast lesions in a more realistic concept of augmented breasts.
5.3. **X-ray imaging simulation**

X-ray projection images from the developed software phantoms were simulated with the Monte Carlo module of the XRayImagingSimulator (Bliznakova et al 2010), at different incident energies between 20 keV and 30 keV.

5.3.1. **Simulated Imaging protocols**

The mammography simulation included generation of six mammography images from each phantom with source to object and source to detector distances of 600 mm and 650 mm, respectively. The incident
photon flux was $5 \times 10^5$ photons/pixel, while the detector response was not simulated. For the simulation study with the step-wedge and the phantom with the realistic implant shape, projection images with a size of 770 pixels×770 pixels and a pixel dimension of 140 μm×140 μm were generated. For the simulation study with the snail phantom, the images had a size of 400 pixels×400 pixels and pixel dimensions of 100 μm×100 μm, generated at a source to detector distance of 660 mm. Breast tomosynthesis protocol included generation of projection images in an isocentric mode, with the x-ray source and the detector both rotating around the phantom with the same direction. The acquisition parameters, i.e. distances, image size and resolution, the total photon flux as well as the incident beam energies, were kept the same. Specifically, for the step-wedge phantom, 31 projection images within an acquisition arc of 60° were generated. For the snail phantom 21 projections within an acquisition arc of 40° were calculated, while for the third phantom 26 projections were generated within an acquisition arc of 50°. In all BT simulations, a 2° increment was used. Imaging protocols were chosen based on the geometrical characteristics of each phantom and a study on the optimal acquisition arcs for BT carried out in advance and presented in the following chapter. Tomosynthesis images were reconstructed with the use of a modified Multiple Projection Algorithm (Kolitsi et al 1992, Malliori et al 2010). All projection images were preprocessed with a ramp filter applied prior to reconstruction. A pseudo-3D representation of each phantom was produced from the reconstructed axial planes at every 1 mm of the total volume.

5.3.2. Optimal Parameters for BT acquisition arcs

Initially, the optimal acquisition arc and number of projections were investigated by setting a simulation study with the use of the snail phantom. Three different sets of images from this phantom were generated at 24 keV by setting the source to the center of rotation distance to 600 mm and the source to detector distance to 660 mm. In addition, the total incident photon fluence (and thus the incident dose) was kept equal to the case of generating a single projection image. The first tomosynthetic volume consisted of 21 projections, each one simulated with approximately $24 \times 10^3$ photons/pix for each projection in an arc of -20:2:20. The second set of images contained 41 projections, equally generated within an arc of 80°, with 12.195 photons/pix per projection. Finally, the third set of images contained 21 projections simulated within an arc of 20° and each projection image was generated with 23.810 photons/pix. Image quality was evaluated in terms of CNR. By increasing the acquisition arc from 40° to 80°, the z resolution increased and the CNR values were slightly better. On the other hand, the blurring of the out of plane objects increased, thus resulting in lots of artifacts. By reducing the arc from 40° to 20°, the CNR reduced dramatically, especially for high thicknesses of implant, where it was almost impossible to detect the μCs.
In addition, the artifact spread function (ASF) (Zhang et al 2006) was used to quantify the “out of plane” artifacts observed in planes outside the plane of focus.

Five μCs from different planes in z direction have been chosen for ASF calculation. These were the μCs placed under 1, 3, 7, 10 and 13mm of implant material. To calculate the ASF, the intensity of in focus and out of focus ROIs of the μCs has been used according to Equation 3.5. The ROI size for mean pixel intensities of the feature and background was chosen to be 4 pixels and 102 pixels, respectively. A comparison of ASF between the three different acquisition geometries was made.

As it can be seen in Fig. 5.2, the ASF was wider for the 20° arc than for the 40° and the 80°. In addition, between 40° and 80° there were no significant changes in the ASF. As a result, the increase of the angular range reduces the artifacts in adjacent planes in the case of small high contrast features. The above results showed that there is an upper limit of acquisition arc above which artifacts cannot be further reduced. The results from this study suggest that acquisition arcs must be within 40° to 80° in case of high contrast features imaging. Therefore, the acquisition arcs of 60° for the step-wedge phantom, 40° for the snail and 50° for the breast-like phantom were chosen.

![Fig. 5.2: ASF for five μCs placed under 1mm, 3mm, 7mm, 10mm and 13mm of silicone gel for three different acquisition arcs at 24keV](image-url)
5.3.3. Monte Carlo code

X-ray interactions in all breast models were simulated using an in-house developed Monte Carlo x-ray simulation software package (Bliznakova et al 2010). This program has been reworked for voxel-based phantoms and adjusted for mammography simulations (Bliznakova et al 2005, Bliznakova et al 2012). Photon transport is modeled with the Monte Carlo method. The program follows the histories of single photons, emitted from the x-ray source, as they pass through the phantom and then reach the detector. The distance between two successive interactions is sampled, based on the relevant attenuation cross-sections, accounting for the different media along the photon path. For the mammographic x-ray energy range used, three interaction processes are considered, i.e. the photoelectric effect, the coherent scattering and the incoherent scattering, as described earlier in Part 1. Each time an interaction occurs inside the phantom, one of the three interaction processes is selected by random sampling, according to the relative cross-sections of the processes at the specific photon energy and medium (Lazos et al 2003). To speed up the simulations, parallel processing in GPU was adopted.

5.3.4. Image quality evaluation metrics

The visibility of breast lesions was assessed in terms of CNR, defined in Chapter 3 from Equation 3.1.

For the simulations using the step-wedge phantom, the average value $I_{obj}$ was calculated over a rectangle region of interest (ROI) of 6 pixels inside each detectable μC, while the average background value $I_{back}$ was estimated over a rectangle area of 176 pixels in the corresponding silicone block.

For the simulations using the snail phantom, the same methodology as in the CNR measurements was followed. The average value $I_{obj}$ was calculated over a rectangle ROI of 4 pixels inside each detectable microcalcification and the average background value $I_{back}$ was calculated for a square-like region of 102 pixels surrounding each μCs.

For the images obtained from the third phantom, rectangular ROIs were considered inside each feature for measuring the average value, while the average background value $I_{back}$ was calculated based on an area of 126 mm×70 mm, selected nearby the implant between the group of masses and the group of μCs.

The ability to visualize low-contrast, subtle objects called contrast resolution has been used. According to the Rose model (Rose 1948), to reliably identify an object, the signal to noise ratio should be higher than
5. In our work, a relative variation of the CNR of each breast lesion was used to predict its detectability. In addition, the visibility was evaluated by three experts working in the field of medical image processing. They were asked to indicate if breast lesions were visible in the images or not. In this way, a minimum CNR was set equal to 2 for high contrast feature detection.

For the stochastic error of our results, each simulation was repeated four times. In all cases, the standard deviation was not larger than 2%. As a result, error bars were not shown in figures for better visualization of the results.

5.4. Results in the case of mammographic images

Six mammographic images for each phantom were generated using monochromatic beams at 20, 22, 24, 26, 28 and 30 keV.

5.4.1. Step-wedge phantom

Regions of interest from mammographic images of the first phantom and line profiles taken horizontally across these images are shown in Fig. 5.3. Visual assessment of those images shows that the detectability of μCs is improved when increasing the incident beam energy from 20 keV to 30 keV. More steps of the phantom can be distinguished (Fig. 5.3(a)) and as a result μCs located under thicker implants are visualized. Specifically for beam energy of 20keV, μCs covering up to 12 mm of implant thickness can be detected. For beam energy of 30 keV, the upper thickness limit of detectability increases to 22 mm of implant material. This improvement in μCs visualization can also be seen from the line profiles shown in Fig. 5.3(b), where the y-axis represents the intensity values.

The results from the visual assessment were well confirmed by the CNR evaluation, shown in Fig. 5.4a. In this figure, a comparison of underlying μC CNR values, calculated for the studied incident energies versus the overlapped implant thickness, is shown. An upper implant thickness limit for feature detection was established by setting a threshold of CNR equal to 2. In detail and according to Fig. 5.4a, at 20 keV up to 12 mm of implant thickness can be detected, at 22 keV up to 16 mm, at 24 keV up to 18 mm, at 26 keV and 28 keV up to 20 mm and at 30 keV up to 22 mm. For thinner implants, lower energies are optimal for irradiation since they result in better CNR. On the contrary, for thicker implants, higher energies are
optimal for high contrast feature detection. Finally, for implants with a thickness greater than 22mm, μCs were not visible on any of the simulated mammography images.

Fig. 5.3: Comparison of (a) the six mammographic images acquired at different energies from the step-wedge phantom and (b) the corresponding line profiles of each image taken across the white line as illustrated in the first image of 20keV.
Considering the results presented with the use of the step-wedge phantom, an identical set of experiments was performed with the use of this phantom but with the silicone step absence. Subsequently, it was made possible to investigate the alterations in the optimal energies for the detection of $\mu$Cs and their CNR values when different thicknesses of silicone are present. The results of these experiments are illustrated in Fig. 5.4c. This figure, compared with Fig. 5.4a, shows the dramatic reduction on the CNR values for the detection of $\mu$Cs as a result of the silicone presence. This reduction is in the order of 45% for a silicone overlap of 2 mm and increases with the increase of silicone thickness. Furthermore, the optimal energies for the detection of $\mu$Cs alter with the silicone slabs insertion depending on their thicknesses. Our results are in agreement with (Bernhardt et al 2006), stating that for the detection of 0.2mm $\mu$C in a block of breast tissue 50% adipose and 50% glandular 4cm thick, the optimal energy appears between 20 keV and 22 keV. The results of this experiment clearly show how a silicone insertion affects the image quality of a mammogram and how those images lack in contrast depending on the thickness of silicone gel.

5.4.2. Snail phantom

Mammographic images acquired with the snail phantom under the same conditions as described previously are presented in Fig. 5.5(a). The contrast and the brightness in each image were set to values that allowed the visualization of the $\mu$Cs under the thickest implant slab. A line profile, taken across each image passing over seven $\mu$Cs placed under implant slabs with thicknesses between 7mm and 13mm, is
shown in Fig. 5.5(b). The calculated CNR for the total of μCs under different implant depths in the energy interval 20-30 keV is presented in Fig. 5.6a. Visual assessment as well as line profiles and CNR values showed slightly higher upper limits of detectability compared with the corresponding values of the step-wedge phantom. This increase is relevant with the increased ESE and the resolution applied in the case of the snail phantom. At 20 keV, μCs overlapped up to 13mm by silicone gel can be detected. As in the previous experiment, increasing the incident energy resulted into improved detection of μCs placed under thicker implants. The maximum slab thickness under which μCs are detectable is 30mm at 30 keV. Detection of μCs placed under thicker implant material (>30mm) was not feasible, despite the use of a phantom with different geometrical characteristics.
Fig. 5.5: Comparison of μCs visibility at different incident energies simulated from the snail phantom: (a) the six mammographic images acquired at different energies from the snail phantom and (b) the corresponding line profiles of each image taken across the white line as illustrated in the first image of 20keV.
5.4.3. Phantom with a realistic silicone gel implant shape

The use of this phantom allowed evaluating the influence of the keV on the visibility of both low contrast features, like breast masses, and high contrast features, like μCs, appearing on the generated mammograms. Figure 5.7 shows simulated mammographic images of this phantom under the same initial conditions as described in the previous experiments. The inserts in Fig. 5.7 depict the area of the image where the μCs C_5 and C_6 exist. As it can be seen, at lower energies these two μCs in the silicone texture were not visualized (Fig. 5.7(a) and Fig. 5.7(b)), in contrast with the images simulated with higher incident energies (Fig. 5.7(e) and Fig. 5.7(f)), where these two μCs are pictured. The calculated CNR values for six μCs (C_1-C_6) adjacent to silicone as well as for the three masses (M_1-M_3) are listed in Table 5.1. Furthermore, calculations on the total area of the phantom corresponding to 234,855 pixels show that 75.8% of the breast equivalent background could be imaged, while a 24.2% that corresponds to the implant is obscured. On this point, it should be mentioned that there are no masses positioned under the silicone, as they could not be detected at any imaging condition, unlike μCs, due to their inherent high contrast.
Fig. 5.7: Simulated mammographic images of the phantom with realistic silicone gel implant. The inserts show the area where the two μCs which are covered from the implant should appear at different energies (a) 20keV (b) 22keV (c) 24keV (d) 26keV (e) 28keV and (f) 30keV.

Table 5.1: CNR values for selected breast features (Fig. 5.7(a)) that appear on the simulated mammographic images.

<table>
<thead>
<tr>
<th>Feature/Energy</th>
<th>20 keV</th>
<th>22 keV</th>
<th>24 keV</th>
<th>26 keV</th>
<th>28 keV</th>
<th>30 keV</th>
</tr>
</thead>
<tbody>
<tr>
<td>M1</td>
<td>14.2</td>
<td>13.2</td>
<td>9.7</td>
<td>6.2</td>
<td>3.9</td>
<td>2.5</td>
</tr>
<tr>
<td>M2</td>
<td>11.8</td>
<td>10.4</td>
<td>6.7</td>
<td>3.3</td>
<td>1.0</td>
<td>0.3</td>
</tr>
<tr>
<td>M3</td>
<td>10.4</td>
<td>8.9</td>
<td>5.4</td>
<td>1.9</td>
<td>1.1</td>
<td>0.4</td>
</tr>
<tr>
<td>C1</td>
<td>106.8</td>
<td>103.8</td>
<td>78.4</td>
<td>50.3</td>
<td>31.6</td>
<td>20.5</td>
</tr>
<tr>
<td>C2</td>
<td>109.5</td>
<td>106.4</td>
<td>79.6</td>
<td>50.6</td>
<td>31.2</td>
<td>19.8</td>
</tr>
<tr>
<td>C3</td>
<td>71.1</td>
<td>66.5</td>
<td>48.2</td>
<td>30.7</td>
<td>18.9</td>
<td>12.3</td>
</tr>
<tr>
<td>C4</td>
<td>72.4</td>
<td>67.8</td>
<td>50.2</td>
<td>32.1</td>
<td>20.7</td>
<td>14.2</td>
</tr>
<tr>
<td>C5</td>
<td>0.5</td>
<td>0.6</td>
<td>0.7</td>
<td>0.8</td>
<td>0.9</td>
<td>1.1</td>
</tr>
<tr>
<td>C6</td>
<td>0.6</td>
<td>0.7</td>
<td>0.8</td>
<td>1.0</td>
<td>1.1</td>
<td>1.1</td>
</tr>
</tbody>
</table>

Table 5.1 reveals that CNR values for findings C5 and C6 in the silicone texture increase with the increase of incident energy, in contrast with the rest of the findings, adjacent to silicone in the breast tissue texture,
which decrease. This, however, does not influence the detection of high contrast features, since the values are greater than the detection threshold (set to \( \text{CNR}_{\text{min}} = 2 \)). On the other hand, this decrease turns out to be critical for the detection of low contrast features, like breast masses. As it can be seen from the image in Fig. 5.7(f), generated for 30 keV, both clusters of \( \mu \text{Cs} \) are detected and clearly visualized, while the small mass (M3) with a radius of 1mm is characterized of reduced sharpness and unclear shape.

5.5. Results in the case of tomosynthetic images

Similarly to the previous examination of mammographic images, a BT investigation was carried out. Six tomosynthetic volumes for each phantom were generated using monochromatic beams at 20, 22, 24, 26, 28 and 30 keV.

5.5.1. Step-wedge phantom

Regions of interest taken from reconstruction planes with \( \mu \text{Cs} \) on focus at 20, 22, 24, 26, 28 and 30 keV of the first phantom shown in Fig. 5.8 were evaluated. The first column of Fig. 5.8 illustrates ROIs for the \( \mu \text{C} \) under the thinnest slab of implant (2mm), the second one ROIs for the last \( \mu \text{C} \) that is detected and the third one ROIs for the \( \mu \text{C} \) under the thickest slab of implant (36mm). Plots of calculated CNR values for the six incident energies are shown in Fig. 5.4b. As it was expected and previously shown for mammographic images, visual assessment and CNR values showed that increasing the energy of the incident x-ray beam results in improved detection of \( \mu \text{Cs} \). Quantitatively, a beam with energy of 20 keV may result to the detection of \( \mu \text{Cs} \) placed under an implant with a thickness of up to 8mm, while 30 keV beam energy results in the detection of \( \mu \text{Cs} \) up to 14 mm. Those limits of implant thickness are lower compared with the corresponding mammographic case. The optimal energies for the detection of \( \mu \text{Cs} \) under most of silicone gel slabs were found to be 28 keV and 30 keV based on the CNR values. Besides, BT showed the ability to detect \( \mu \text{Cs} \) under the thickest block of 36mm even at lower energies, but in the region 14 mm to 35 mm the detectability was not feasible. This is a result of the x-ray source movement combined with the geometrical characteristics of the phantom that will be evaluated later on.
5.5.2. Snail phantom

The BT slices obtained from images simulated with the snail phantom resulted in lower CNR values compared to the mammography projection images. However, BT allowed to visually detect the breast...
lesion under the implant slab of 49 mm. Fig. 5.6b presents the calculated CNR values versus the implant thickness at different energies. This plot shows a peculiar CNR behavior compared to the step-wedge phantom. After 12 mm of implant thickness (which was the mean upper limit of detectability for the previous phantom), the graph is characterized by a pattern of local maxima and minima which reminds of a damped oscillation. The characteristic of the local maxima is that they correspond to the CNR of $\mu$Cs under almost each corner slab of the phantom with thicknesses of 19, 24, 29, 33, 37, 40, 43, 47 mm. Moreover, according to the graph of CNR for thicknesses above 32 mm, all the $\mu$Cs are detectable. This result is another example of the x-ray source movement combined with the geometrical characteristics of the phantom.

5.5.3. **Phantom with a realistic shape of silicone gel implant**

Similarly to previous sections, ROIs taken from 2 reconstruction planes at 20 keV to 30 keV of the third phantom are shown in Fig. 5.9. In Fig. 5.9a and 5.9b, the BT planes where the big cluster of $\mu$Cs and the masses with the small cluster of $\mu$Cs are respectively illustrated. Visual assessments and calculations on those ROIs showed that BT images were able to image a larger breast equivalent area. Specifically, the BT slice where the big cluster of $\mu$Cs is in focus demonstrated an increase of 14,644 pixels in the visualized breast area compared to the corresponding mammographic projection images (Fig. 5.7). A result of this increase is the segregation of the fourth $\mu$C from the silicone texture and its clear visualization (Fig. 5.9(a)). Calculated CNR values of the simulated breast lesions are listed in Table 5.2. In addition, BT resulted in improved visualization for the two $\mu$Cs $C_5$ and $C_6$ hidden under the implant. Moreover, $\mu$Cs and masses in the breast tissue background were best visualized at incident energies of 22-24 keV and 20-22 keV respectively, while $\mu$Cs in the silicone background were best visualized at 30 keV based on CNR values. At this energy, the whole cluster of $\mu$Cs was completely visualized. However, as shown in these images, low contrast features are characterized with slightly reduced visibility. Breast tomosynthesis shows advantages in visualizing low contrast masses of small size ($M_3$) over mammography in high energies. As it was expected, the CNR of low contrast features is lower for the case of the ramp pre-filtering. Ramp is a high pass filter that can enhance the edges and increase contrast but also allows the high frequency noise in the image to remain.
Fig. 5.9: Cropped images (a) from the plane where the cluster of μCs with a radius of 0.4 mm is in focus and (b) from the plane where the small cluster and the three masses are in focus.

Table 5.2: CNR values for the main features (Fig. 5.7(a)) of the BT images acquired from the breast-like phantom.

<table>
<thead>
<tr>
<th>Feature/Energy</th>
<th>20</th>
<th>22</th>
<th>24</th>
<th>26</th>
<th>28</th>
<th>30</th>
</tr>
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<tr>
<td>M1</td>
<td>1.3</td>
<td>1.2</td>
<td>1.1</td>
<td>1.0</td>
<td>0.9</td>
<td>0.8</td>
</tr>
<tr>
<td>M2</td>
<td>0.8</td>
<td>0.7</td>
<td>0.6</td>
<td>0.6</td>
<td>0.5</td>
<td>0.5</td>
</tr>
<tr>
<td>M3</td>
<td>0.5</td>
<td>0.6</td>
<td>0.6</td>
<td>0.5</td>
<td>0.4</td>
<td>0.4</td>
</tr>
<tr>
<td>C1</td>
<td>26.3</td>
<td>29.5</td>
<td>31.6</td>
<td>29.8</td>
<td>28.6</td>
<td>26.9</td>
</tr>
<tr>
<td>C2</td>
<td>27.2</td>
<td>30.6</td>
<td>32.4</td>
<td>31.5</td>
<td>30.2</td>
<td>28.3</td>
</tr>
<tr>
<td>C3</td>
<td>24.9</td>
<td>25.1</td>
<td>22.8</td>
<td>23.5</td>
<td>20.3</td>
<td>19.3</td>
</tr>
<tr>
<td>C4</td>
<td>14.4</td>
<td>14.4</td>
<td>14.1</td>
<td>14.9</td>
<td>11.7</td>
<td>13.6</td>
</tr>
<tr>
<td>C5</td>
<td>0.04</td>
<td>0.32</td>
<td>0.84</td>
<td>1.59</td>
<td>1.94</td>
<td>2.67</td>
</tr>
<tr>
<td>C6</td>
<td>0.61</td>
<td>0.88</td>
<td>1.18</td>
<td>1.97</td>
<td>2.81</td>
<td>3.25</td>
</tr>
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</table>
5.6. Performance of Mammography versus BT

The comparison between the two modalities assessed at the same photon fluence of $5 \times 10^5$ photons per pixel in this study showed that mammography resulted in higher upper thickness limits of detectability for high contrast features of small size. Specifically, the average limits for the mammography case are 12.5 mm at 20 keV, 16.5 mm at 22 keV, 20.5 mm at 24 keV, 21.5 mm at 26 keV, 23.5 at 28 keV and 26 mm at 30 keV. On the contrary, these limits are significantly lower in the case of BT, i.e. 8, 8, 10, 12, 14, and 14 mm, respectively. In Fig. 5.10, comparative graphs of CNR between mammography and BT for both escalating phantoms at different energies are illustrated to facilitate the comparison of the two imaging modalities. Areas of the phantoms with thicknesses above those limits attenuate the radiation making the detection of $\mu$Cs impossible. Alterations of energy in the interval of 20 keV to 30 keV led to improved detectability of $\mu$Cs, while the opposite stands for breast masses. The comparison between BT and mammographic images showed that the latter resulted in superior CNR and less artifacts for $\mu$Cs detection in most of the cases. Our results are in agreement with (Malliori et al. 2012), stating that low signal lesions larger than 1 mm can be visualized better with BT, while 0.2 mm lesions may be better visualized with digital mammography, when compared at equal doses. In addition, the reduction of CNR is considerably high in the cases of low energies and implant thicknesses up to 10 mm, thus making BT an inappropriate modality for this case.

According to the present study, the main advantage of BT over mammography that may not be neglected, apart from the in depth information, is the detection of high contrast features under very thick implant material, which mammography fails to visualize. In BT mode, the x-ray source moves in an arc, in contrast with mammography, where it remains stable, thus resulting in smaller mean free paths for some of the features at some angles. This leads to favorable viewing angles for the visualization of findings overlapped by significant thicknesses of silicone. Characteristic examples are the $\mu$Cs at the corners of the snail phantom that appear in the BT images with higher CNR compared with the rest of this area (Fig. 5.6b), but also the detection of the $\mu$Cs with thicknesses between 30 mm-49 mm of silicone gel that were impossible to be detected in the mammographic images. Even if this phenomenon is related to the phantom geometry and the acquisition arc of the x-ray source, further studies in this direction may confirm that BT is a useful modality for thick implants when the appropriate acquisition arc is used.

Another advantage is the larger visualized breast area (not obscured by the silicone gel) observed in the BT images of the third phantom compared with the mammographic ones. In BT slices (Fig. 5.9(a)), the obscured breast area covers only 17.7% of the image, while in the mammographic image (Fig. 5.7) the obscured area reaches 24.2%. This increased visualization may be critical, since it could reveal important
information. Such an example was the detection of the μC near and under the implant edges that could be clearly visualized in BT, in contrast with the planar image, where the same μC was hidden in the silicone texture. Finally BT demonstrated a superior visualization of small masses adjacent to silicone gel compared with mammography as in the cases of non augmented breasts (Teertstra et al 2010, Gennaro et al 2010).

Fig. 5.10: Comparative CNR graphs for mammography and BT at different energies (a) for the step-wedge phantom and (b) for the snail phantom.

5.7. Conclusion

This study evaluated the quality of both high contrast features, like μCs, and low contrast features, like breast masses, on simulated mammography and BT images in the presence of silicone gel implant of different thickness at beam energies in the interval of 20-30 keV. Lesions placed under thick silicone gel material (>26mm) could not be visualized either with mammography or BT, revealing the problems arising from the implant insertion in breast imaging. Conventional mammographic images resulted in superior CNR values compared to those calculated for tomosynthesis. This is especially valid for μCs
lying under thin implants imaged with low beam energies. Mammography also resulted in higher limits of silicone gel thickness for μCs detection compared with BT.

At low energies (20-22 keV), CNR measurements showed high dependence on implant thickness, while the response of CNR rapidly decreases as the implant thickness increases. This dependence of CNR on the implant thickness becomes less steep with the increase of energy. Alterations of energy in the interval of 20 keV to 30 keV led to improved detectability of μCs overlapped by silicone, while the opposite stands for breast masses. Although BT images resulted in inferior image quality in terms of CNR, they demonstrated an advantage in visualizing a larger breast area and small low contrast lesions because of non-overlaps. Furthermore, BT results showed interesting behavior for certain heights in the snail and the step-wedge phantom, which are attributed to preferable viewing angles. This indicates that the acquisition arc selected in an application where breast screening with silicone presence is needed, is critical. In addition, in some cases the μCs lying under high thicknesses of silicone were detected in BT images, indicating that BT presents some advantages compared to mammography.

Finally, three complicated breast phantoms with silicone gel insertions were designed for x-ray imaging studies. Since no such phantoms are commercially available, their future realization will be an essential tool for image quality assessment of mammograms, optimization of imaging protocols including breast tomosynthesis, and dosimetry, in the case of implant presence in breast.
6. **COMPEX BREAST PHANTOM WITH SILICONE GEL**

Chapter 5 presents a detailed study of the effect that comes along with silicone insertion on the detection of breast lesions. The phantoms introduced for this study had a homogeneous background equivalent to 50% glandular and 50% adipose tissue. In this chapter, the detection of breast lesions in the neighborhood of a silicone gel insertion within a heterogeneous texture is investigated. For this reason, a 3D software breast model with realistic breast tissue distribution is introduced. The heterogeneous background will help to mimic the breast in a more realistic way.

### 6.1. Complex phantom

The complex breast phantom was built with the use of the Breast Simulator, an extension of the XrayImagingSimulator (Bliznakova et al 2010), and is composed of skin, glandular and adipose tissue

![Fig. 6.1: A projection image of the complex phantom with all the individual parts of which it consists.](image)

(Fig. 6.1). The voxel size of the 3D breast model was 0.1 mm in each direction, whereas the whole breast volume was placed in a 3D matrix with size 34.8mm width, 49.3mm length and 30 mm height. The glandular portion occupied 37% of the breast volume. The final breast volume was a compressed version of the abovementioned volume with the use of a compression algorithm (Zyganitidis et al 2007). The compression plates in the Breast Simulator were placed in a way to generate 2cm compressed thickness.
The densities were $p_{\text{gland}}=1.04 \ \text{g/cm}^3$ and $p_{\text{adipose}}=0.93 \ \text{g/cm}^3$ consisting a mixture whose weight of glandular and adipose tissue was 40% and 60% respectively (Hammerstein et al. 1979). Moreover, a silicone gel with a semiellipsoid form and dimensions $14\times21\times8 \ \text{mm}^3$ mimicking a breast implant, as described in Chapter 4.4, two clusters of $\mu$Cs ($\text{CaCO}_3$ with $p=2.8 \ \text{g/cm}^3$) with 0.3 mm and 0.2 mm radius and a water mass with irregular margins were inserted into the 3D breast model. The mass as well as the two clusters of $\mu$C were placed in the closed neighborhood of the silicone gel and some of them overlap. Figure 6.1 presents a projection image at 20 keV, where the breast tissue with the abnormalities and the silicone gel arrangement is depicted.

6.2. Mammographic examination

Two mammographic images 700x700 square pixels with resolution of 100$\mu$m were generated with an incident photon flux $5\times10^5$ photons/pixel at 20 keV with the Monte Carlo irradiation module of the XRayImagingSimulator (Bliznakova et al. 2010). The SDD and SID were 660 mm and 600 mm respectively. The first image was generated with the use of the complex phantom presented in Fig. 6.1. In order to acquire the second image the silicone gel is removed from the complex phantom. The purpose of this experiment is to understand how the silicone gel affects the detection of the main mammographic findings. The 2D projection images acquired with and without the silicone gel can be seen in Fig. 6.2a) and b) respectively.
The detection of mammographic features in all images was assessed visually and quantitatively. Visual inspection of those images acquired with and without the implant (Fig. 6.2) shows how the silicone insertion obscures partially to totally lesions of the phantom. Even with different windowing of the images, some of the lesions were impossible to be detected in the case of silicone overlap. In order to quantify this drawback, an evaluation metric was introduced in terms of CNR (Equation 3.1). In Fig. 6.3, the ROIs under investigation are illustrated. The ROIs 1 to 6 correspond to the big cluster of $\mu$Cs, 7-12 to the small cluster of $\mu$Cs, and ROI 13 to the mass. Regions of interest 14 and 15 are used as a background in the case of $\mu$Cs overlapping with silicone and not, respectively, while ROI 16 is used as a background for the mass.

Fig. 6.2: Mammographic images at 20 keV a) with the implant insertion and b) without.
Table 6.1 presents the resulting CNR values for the total of the main mammographic findings for the complex breast phantom with and without the silicone gel.

Table 6.1: CNR values for the ROIs presented in Fig. 6.3 from a mammographic image acquired at 20 keV.

<table>
<thead>
<tr>
<th>ROI</th>
<th>Area</th>
<th>CNR with Implant</th>
<th>CNR No Implant</th>
</tr>
</thead>
<tbody>
<tr>
<td>CaCo3-1</td>
<td>1</td>
<td>2.28</td>
<td>27.36</td>
</tr>
<tr>
<td>CaCo3-2</td>
<td>2</td>
<td>2.38</td>
<td>27.18</td>
</tr>
<tr>
<td>CaCo3-3</td>
<td>3</td>
<td>0.33</td>
<td>26.65</td>
</tr>
<tr>
<td>CaCo3-4</td>
<td>4</td>
<td>23.61</td>
<td>27.38</td>
</tr>
<tr>
<td>CaCo3-5</td>
<td>5</td>
<td>11.06</td>
<td>26.54</td>
</tr>
<tr>
<td>CaCo3-6</td>
<td>6</td>
<td>0.48</td>
<td>27.17</td>
</tr>
<tr>
<td>CaCo3-s1</td>
<td>7</td>
<td>6.28</td>
<td>22.23</td>
</tr>
<tr>
<td>CaCo3-s2</td>
<td>8</td>
<td>15.44</td>
<td>23.05</td>
</tr>
<tr>
<td>CaCo3-s3</td>
<td>9</td>
<td>22.88</td>
<td>22.37</td>
</tr>
<tr>
<td>CaCo3-s4</td>
<td>10</td>
<td>23.59</td>
<td>23.18</td>
</tr>
<tr>
<td>CaCo3-s5</td>
<td>11</td>
<td>22.14</td>
<td>21.84</td>
</tr>
<tr>
<td>CaCo3-s6</td>
<td>12</td>
<td>23.86</td>
<td>23.5</td>
</tr>
<tr>
<td>Mass</td>
<td>13</td>
<td>1.37</td>
<td>1.4</td>
</tr>
</tbody>
</table>

As it can be seen in Table 6.1, for the big cluster of μCs (ROIs 1-6) that is totally overlapped by the silicone gel, there is a reduction of the CNR, reaching a value 80 times lower. For the smaller cluster (ROIs 7-12), where silicone overlaps only two out of six μCs (7 and 8), a reduction in CNR occurs but with minor differences. The same result occurs for the mass (ROI 13) placed in the neighborhood of the
silicone, where the CNR reduces from 1.4 for the case without the implant to 1.37 with the implant insertion. These results demonstrate that a silicone gel dramatically reduces the CNR values for the case of overlapping but also reduces the CNR values in its neighborhood with breast tissue background.

6.3. Tomosynthesis examination

After the mammography investigation presented previously, tomosynthesis as second imaging modality with the use of the complex phantom is examined. For this purpose, 21 projections acquired at 2° increments in an acquisition arc of 40°, were generated. Tomosynthesis images 700×700 pixels with 100\(\mu\)m resolution of the breast phantom with the implant were obtained by reconstructing projection images preprocessed with a ramp filter, using MPA. The SDD and SID remained the same as in the mammography mode, at 660 mm and 600 mm respectively. Monochromatic incident beams at 20 keV and 24 keV were simulated with the Monte Carlo irradiation module of the XRayImagingSimulator. The incident photon flux was 23.810 photons/proj*pixel resulting in the same total photon flux as in the case of mammographic images. In Figure 6.3, images from the planes where the small and big cluster of \(\mu\)Cs are in focus for the energies 20 keV and 24 keV are presented.
Comparing visually the images acquired with mammography and tomosynthesis mode at 20 keV (Fig. 6.2a and Fig. 6.3), there is an advantage in detectability of μCs in the second case. The overlapping effect is less in the case of BT thus making the detectability of μCs easier. More breast tissue is visualized revealing lesions in the planes where they are focused. Characteristic are the examples of μCs 5, 4, 7 and 8 that are clearly depicted in Fig. 6.3 (first row), in contrast with Fig. 6.2a, where they are obscured.
Moreover, the contrast is changed due to filtration in the case of BT resulting in improved visibility of μCs. As it was expected from previous experiments, the increase of the incident energy from 20 keV to 24 keV improved even more the visibility of lesions that overlap with silicone gel. As in the case of mammography, a quantitative analysis is performed by using the same ROIs. The results of CNR for the case of 20 keV and 24 keV, respectively, are presented in Table 6.2 below.

Table 6.2: CNR for the ROIs of tomosynthesis images of the first phantom with the implant at 20keV and 24 keV.

<table>
<thead>
<tr>
<th>ROI</th>
<th>Area</th>
<th>CNR 20 keV</th>
<th>CNR 24 keV</th>
</tr>
</thead>
<tbody>
<tr>
<td>CaCo3-1</td>
<td>1</td>
<td>1.32</td>
<td>12.54</td>
</tr>
<tr>
<td>CaCo3-2</td>
<td>2</td>
<td>0.85</td>
<td>6.64</td>
</tr>
<tr>
<td>CaCo3-3</td>
<td>3</td>
<td>1.92</td>
<td>14.2</td>
</tr>
<tr>
<td>CaCo3-4</td>
<td>4</td>
<td>27.05</td>
<td>59.1</td>
</tr>
<tr>
<td>CaCo3-5</td>
<td>5</td>
<td>25.03</td>
<td>54.59</td>
</tr>
<tr>
<td>CaCo3-6</td>
<td>6</td>
<td>4.74</td>
<td>21.27</td>
</tr>
<tr>
<td>CaCo3-s1</td>
<td>7</td>
<td>10.17</td>
<td>11.72</td>
</tr>
<tr>
<td>CaCo3-s2</td>
<td>8</td>
<td>14.8</td>
<td>17.48</td>
</tr>
<tr>
<td>CaCo3-s3</td>
<td>9</td>
<td>46.13</td>
<td>37.82</td>
</tr>
<tr>
<td>CaCo3-s4</td>
<td>10</td>
<td>50.81</td>
<td>41.97</td>
</tr>
<tr>
<td>CaCo3-s5</td>
<td>11</td>
<td>47.85</td>
<td>39.27</td>
</tr>
<tr>
<td>CaCo3-s6</td>
<td>12</td>
<td>52.22</td>
<td>44.86</td>
</tr>
<tr>
<td>Mass</td>
<td>13</td>
<td>0.2/2.1</td>
<td>0.17/1.9</td>
</tr>
</tbody>
</table>

Conclusion

In Fig. 6.4 the aggregate results from the experiments with the complex phantom are presented. Table 6.1 as well as Fig. 6.2 show the clear disadvantage in the detectability of lesions and the lack of CNR due to silicone placement in the heterogeneous breast tissue. The values of CNR for high and low contrast lesions drop in the presence of silicone. Specifically, the mean CNR values for the big cluster, the small cluster and the mass decrease from 27 to 6.7, from 22 to 19 and from 1.4 to 1.37, respectively. This decrease of CNR was higher for the lesions that overlap with silicone gel. In most of the cases, BT demonstrates higher values of CNR compared with mammography. This is especially the case for μCs close to the borders of the silicone and in its neighborhood with no overlaps in the in focus planes, except from μCs 1 and 2, where mammography resulted in higher values of CNR. This superiority of mammography for those two μCs (1 and 2 in Fig. 6.3) is due to the high thickness of silicone overlap, measured to be 13 mm and 15 mm, in contrast with the rest that are covered from silicone thickness varying from 4 mm to 9 mm. Furthermore, mammography resulted in higher CNR value compared with BT for the mass. This is a result of the Ramp-filter that has been used mostly for better imaging of the μCs groups. Reconstructed planes without filtration resulted in higher values (2.1 for 20 keV) compared
with mammography for the CNR of the mass. Thus making BT a better modality for the detection of breast masses when no pre-filter was used.

A comparison between the results with the complex phantom and the results from previous chapters with homogeneous background phantoms shows a reduction in CNR values for lesions in the first case. This reduction is a result of the complexity of the base material where the lesions were inserted. Moreover, this reduction might not be so critical for the detection of μCs but for the mass, results are disappointing. As it can be seen in Fig. 6.2a, the complexity of the background in correlation with the scatter of silicone's neighborhood makes it difficult to detect the mass even if its position was known a priori. Another reason that makes the detection of the mass even harder is that half of it is totally obscured because it overlaps with silicone.

![CNR values for the mammographic features of the complex phantom with and without silicone gel insertion in BT and mammography mode.](image)

Fig. 6.4: CNR values for the mammographic features of the complex phantom with and without silicone gel insertion in BT and mammography mode.

Another factor influencing the detectability of lesions apart from the complexity of the background is its density. The complex phantom presented here had a heterogeneous background of 40% glandular and 60% adipose tissue and this density makes the detectability of masses difficult.

86
6.4. Further Investigation

In previous experiments it was demonstrated how silicone dramatically affects the visibility of lesions in case of overlapping. In addition, a CNR reduction of lesions in the neighborhood of silicone is also observed. This may not be influential in the detection of high contrast features, like \( \mu \text{Cs} \), because of their inherent high contrast, but for low contrast features like breast masses it is crucial. In order to understand how the neighborhood of silicone can be affected by silicone's existence, two sets of ROIs were arranged and positioned at different distances from the center of the silicone insertion. The mean values of these ROIs are calculated in projection images acquired with and without the implant. For this experiment a new complex phantom was used with less dense background and the characteristics presented below. The breast phantom illustrated in Fig.6.5 which is actually under compression of 2cm constitutes from:

- Breast tissue (background) – adipose 80% and gland 20%. The bin file has width: 212,76mm, length: 156,6mm and height: 83,43mm.
- Silicone Gel Implant \((CH\text{3}[Si(CH\text{3})O]_2Si(CH\text{3})_3)\). A semi-ellipsoid with dimensions \((72, 76, 30)\)mm placed at the base of the breast phantom.
- Three sets of microcalcifications (\(\text{CaCO}_3\)) with 0,3mm, 0,4mm and 0,5mm radius respectively.
- An abnormality almost at the center of the breast phantom.

Fig. 6.5: A projection image of the complex phantom (80% adipose-20% glandular tissue) with all the individual parts of which it consists.
According to Fig. 6.6, there is a reduction in the mean value of ROIs in the case of silicone gel insertion. This reduction is essential in areas close to the silicone gel, whereas it is less significant as we move far away from it. For example, in Fig. 6.6a) ROIs in the periphery of the implant (1,3,4) have a difference of 20 units in their mean values. On the contrary, ROIs 2, 5 and 6, which are more distant from silicone, present minor differences in their mean values, i.e. 10, 5 and 2 respectively. Similarly, in Fig. 6.6b), the mean values of ROIs (1 to 7) calculated from images with and without the implant present smaller differences as we move away from the implant. This difference is 20 units for ROI 1 placed close to the phantom and becomes almost zero for ROI 7. This phenomenon can be explained due to high absorption and scatter on silicone material that produces a loss of visibility in the areas adjacent to the implant.
Since in the previous chapter the identification and visualization of the mass was difficult because of the semi-overlap with silicone and because of the high density of breast tissue another trial is done with the use of this phantom. In Fig. 6.7 images of the mass from 2D image with and without the implant as well as BT planes with pre-filtering and not are presented.

![Images of the mass at 2D images with and without the implant, and BT planes with the mass in focus with pre-filtering and without.](image)

From the second row of Fig. 6.7 is clear the advantage of BT over mammography to visualize breast masses with low density even though no pre-filtering was applied to enhance its visibility. Moreover from the BT slices the volume of the mass can be calculated but also its irregular margins are better depicted which is very important for the characterization of a lesion. The superiority of BT without pre-filtering over mammography was confirmed from the CNR values presented below.

<table>
<thead>
<tr>
<th>Mass</th>
<th>2D-No_impl</th>
<th>2D-impl</th>
<th>BT-filtered</th>
<th>BT-unfiltered</th>
</tr>
</thead>
<tbody>
<tr>
<td>CNR</td>
<td>4.7</td>
<td>4.6</td>
<td>0.3</td>
<td>6.2</td>
</tr>
</tbody>
</table>

As it can be seen in this table the CNR value increased from 4.7 to 6.2 when BT was used as an imaging modality. This increase that BT provides may be critical for the identification of a low contrast lesion growing into augmented breast tissue.
7. INVESTIGATION OF ALTERNATIVE IMAGING MODALITIES

From the previous experiments it is concluded that beam energies 20 keV-30 keV were not able to detect high contrast features placed under thick (more than 30 mm) silicone gel material. Even BT, which could detect μCs under the thickest slab of implant, resulted in undetectable areas of thicknesses. According to the HVL and average distance for silicone gel, in order for an implant thickness of 40mm-50mm to be penetrated, the beam energy has to be increased significantly. For this reason, the visualization and detectability of features will first be investigated in BT mode with higher beam energies. In order to find a good balance between penetration and contrast, projection images were acquired with the use of the snail and step-wedge phantoms in the energy interval 40 keV to 120 keV. The Monte Carlo module of the XRayImagingSimulator (Bliznakova et al 2010) was used again for the simulations, whereas the simulated imaging protocols were kept the same as they were stated in Chapter 5. For all the energies, the incident photon flux was $5 \times 10^5$ photons/pixel resulting in lower doses for higher energies. In Fig. 7.1 is illustrated the CNR of each μCs of both phantoms in correlation with the thickness of implant material.

In a previous section it was illustrated that for energies 20-30 keV, μCs could be detected when overlapped with up to 20 mm of silicone gel. As it was expected, the use of energies higher than the conventional area of breast screening results in a significant CNR reduction. As it can be seen in Fig. 7.1, the strange pattern with the minimums and maximums disappears and the CNR stabilizes between 2 and 4. Increasing the energy above 40 keV made it possible to detect almost the total of μCs that overlapped with implant material.
**Dual Energy mammography and BT application**

Taking into account the above results, i.e. that higher energies were able to detect the total of $\mu$Cs and that the energy interval 20 keV-30 keV resulted in higher CNR values for thick implant material, the usage of Dual Energy (DE) is going to be investigated. The aim is to examine whether such a modality could result in better visualization of high contrast features such as $\mu$Cs of 0.2 mm radius in case they are overlapped by silicone gel in comparison with standard mammography and BT. There is a variety of different linear and non-linear algorithms for the combination of low and high energy in DE applications. Dual Energy subtraction imaging techniques have been applied to mammography in many theoretical and experimental studies and demonstrated enhanced visualization compared with conventional mammography (Johns et al 1985, Boone 1991, Lemacks et al 2002, Bliznakova et al 2006, Kamarianakis et al 2009, Carton et al 2010). The dual energy images in this study have been calculated as it was described in (Bliznakova et al 2006) by the equation:

$$DE = \log N_H - R \log N_L + k, \quad R = \frac{\mu_H}{\mu_L}$$

Equation 7.1

where $N_H$ and $N_L$ are the photons adsorbed in the detector for the high energy and low energy respectively and $k$ varies with the detector absorption coefficient. In our simulations, the detector is ideal and so $k$ equals to 0. Different combinations of energies have been used in order to find the most optimized one.

**7.1. Dual Energy subtraction mammography**

Firstly simulations of DE subtraction mammography with the use of both escalating phantoms are examined. Different combinations of low and high incident energies in the areas 20 keV to 30 keV and 40keV to 120 keV respectively were tested in order to find the optimal set for better visualization of $\mu$Cs. The same acquisition parameters, photon flux and image sizes were used as in the corresponding simulations of Chapter 5. Moreover, the same ROIs inside each $\mu$Cs were taken into account in order to quantitatively evaluate the results in terms of CNR. For the multiplier factor $R$ of Equation 7.1 the value $R_{CaCO_3} = 0.2645$ was chosen after an extended trial of different values. The CNR for the $\mu$Cs of the snail and the step-wedge phantoms are illustrated in Fig. 7.2 and Fig. 7.3 respectively. In these graphs only the combination of energies that resulted in worth mentioning CNR values are presented.
The CNR values calculated from the acquired DE mammographic images show that for some combinations of low and high energies the detectability of almost all the μCs is possible. In Figure 7.2 the combinations 20-40 keV and 28-40 keV resulted in CNR values adequate (>2) for the detection of almost all the μCs of the snail phantom. In the corresponding simulation of conventional mammography, there was a limitation in the thickness of silicone equal to 30 mm for the detection of μCs. In the case of DE, the CNR, especially for small thicknesses of implant material, dramatically falls but its mean value is
around 3, which is enough for a trained eye to detect high contrast features of small size. This is especially the case for 28-40 keV, where the CNR stays above the limit of detectability up to 45 mm silicone thickness. Interesting CNR results were acquired also from the combination of 20-28 keV, in particular for thicknesses up to 26 mm; however, above this value there is a sudden drop. The visual inspection of the acquired DE mammographic images that was done was in agreement with the abovementioned quantitative evaluation. In Fig. 7.4 a) and b) the two images for the optimal combination of energies 28-40 keV and 20-40 keV are illustrated.

The DE mammographic images acquired with the use of the step-wedge phantom showed similar results with the ones of the snail phantom. The values of CNR drop for small thicknesses compared with the corresponding convention images but for some combinations of energies the total of \( \mu \)Cs could be detected. The limit of silicone thickness under which \( \mu \)Cs could be detected in conventional images was 22 mm (for 30 keV), whereas for DE it reaches the whole block of silicone thickness up to 36 mm. The optimal combination of energies that resulted in adequate values of CNR for detectability in small and high thicknesses is the 20-28keV (Fig. 8.3). In addition, the energies 28-40keV and 20-40 keV resulted in almost the same values of CNR for thicknesses above 22 mm, but there is a significant decrease below this value. In Figure 7.5 a) and b) the two images for the optimal combination of energies 20-28 keV and 28-40 keV are illustrated.
Dual energy subtraction mammography showed an improvement in the detection of high contrast features compared with conventional mammography. For this reason, the use of DE BT is going to be examined with the use of both escalating phantoms. For the simulations the same acquisition parameters, photon flux and image sizes were used as in the corresponding simulations of BT in Chapter 5 to allow comparisons. The same simulations for the optimum combination of energies as in DE mammography were performed. For the multiplier factor $R$ of Equation 7.1, the same value $R_{\text{caco}} = 0.2645$ was used, since it resulted in better images also for the case of DE BT. The CNR for the $\mu$Cs of the step-wedge and snail phantoms that resulted in worth mentioning values are illustrated in Fig. 7.6 and Fig. 7.7 respectively.
The CNR measurements of Fig. 7.6 and Fig. 7.8 as well as visual inspection of the images acquired, showed higher limits of silicone thicknesses compared with conventional BT under which μCs can be detected. Specifically, for the step-wedge phantom this limit was 14 mm (at 30 keV), whereas for the snail phantom there was a problematic zone from 15 mm to 35 mm. In contrast with conventional BT, DE BT improves these limits, as the first one reaches up to 22 mm for the step-wedge and almost the total of μCs can be detected in the snail phantom. Similarly to DE mammography, there is a reduction in CNR values, but – for the optimal combination of energies – this reduction stays above 2, even for high thicknesses. Moreover, as in DE mammography, the combination 28 keV-40 keV resulted in an improved
detectability. But DE BT revealed that the energies 28 keV-60 keV are among the optimal combinations for the snail phantom.

7.3. DE with the use of breast phantom

In order to evaluate the results of the previous experiments and the advantages of DE in a more realistic way, a simulation with the breast-like phantom was run. The use of this phantom also allowed investigation of DE for the detection of low contrast features (masses). The construction of this phantom has been described in a previous section. In a previous study presented in Chapter 5, the visualization of the 2 sets of μCs and the three masses of the phantom was investigated at the presence of silicone gel in the energy interval 20keV-30keV.

![Fig. 7.8: Simulated DE mammographic images of the breast-like phantom with energies sets 28 keV-40 keV, 20 keV-28keV and 20 keV-40 keV from left to right.](image)

The result of this study showed that in both mammography and BT the optimal energy for the detection of the whole big cluster of μCs was 30 keV. However, at this energy masses are characterized with reduced visibility, sharpness and unclear shape. Dual energy mammography and DE BT simulations were run by
keeping constant all the imaging configurations as in the corresponding conventional modalities. Based on the previous results of DE, different combinations of energies and R values were tested. The goal at this point was to examine whether DE may result in a good visualization of both μCs covered by silicone and masses in its neighborhood, in a single image. In Fig. 7.8 DE mammography images for three combinations of low and high energy are illustrated. These images were superior among the total of combinations tried and they were acquired with the multiplier factor equal to $R_{CaCO_3} = 0.2645$.

The image acquired with energies 20 keV-40 keV resulted in an enhanced visualization of both features and superior detectability compared with conventional images. As it can be observed in Fig. 7.8, the total of the big cluster of μCs – even the three that overlapped with silicone – is strongly depicted and the three masses are visualized concurrently. The image acquired with the combination 20 keV-28 keV in Fig. 7.8 achieves a good visualization for the total of μCs but fails in the visualization of masses. Finally, the combination of energies 28 keV-40 keV as well as the rest of combined energies resulted in inferior visualization of both features.

Consequently, simulations of DE BT with the use of the breast-like phantom were run. The optimum sets of energies for the case of DE BT were 20 keV-40 keV, 28 keV-40 keV, and 20 keV-80 keV. The BT planes produced for these sets of energies are illustrated in Fig. 7.9. The first and second column of Fig. 7.9 correspond to the plane where the big cluster of μCs is in focus, named Plane 1, and to the plane where the masses and the small cluster of μCs are in focus, named Plane 2, respectively. The images shown in the last row of the same figure were acquired with the use of $R_2$ equal to 0.4036, whereas the rest with $R_{CaCO_3} = 0.2645$. All four cases of DE BT presented in Fig. 7.9 succeeded to strongly visualize the big cluster of μCs but the set of energies 28 keV-40 keV failed to visualize the group of masses. Visual inspections and comparison of the resulting DE BT images showed that the set of energies 20 keV-40 keV (first row of Fig. 7.9) resulted in the most advantageous visualization of both low and high contrast features. Moreover, DE BT significantly improved the visualization of both masses and μCs in a single acquisition compared with DE and mono-energy mammography. Finally, DE BT of 28 keV-40 keV resulted in an enhanced visualization of both types of lesion and a reduction of noise, thus making the detection of small mass $M_3$ easier compared with mono-energetic BT.
Fig. 7.9: Dual energy tomosynthetic planes of different combinations of the breast-like phantom, where the big cluster of μCs are in focus (plane 1) and the masses with the small cluster of μCs are in focus (plane 2).

Dual Energy BT seems to be an appropriate modality for breast imaging when an implant is inserted and both masses and microcalcifications need to be detected.
7.4. BT performed in a discontinuous acquisition arc

Knowing the structure of the snail phantom and the fact that lesions are located under the silicone gel slabs, an experiment of BT in particular arcs was carried out. The acquisition arc was in a discontinuous area of degrees -80°:2°:-60° and 60°:2°:80°. These intervals have been chosen in order to reduce the mean free path of the x-rays compared with the one at 0°. For example, at 0° the beam has to travel 49 mm of high attenuated material to reach the \( \mu \)C in the center of the phantom under the thickest slab of silicone gel. Breast tomosynthesis protocol included generation of projection images in an isocentric mode, with the x-ray source and the detector moving around the snail phantom in a discontinuous arc -80°:2°:-60° and 60°:2°:80°. Twenty two projection images 600 pix* 600 pix in size were acquired with a resolution of 100\( \mu \)m and incident energy 28 keV. The SID and SDD were 600 mm and 660 mm respectively, as in previous experiments with the same phantom. The number of photons per projection were 22,727 resulting in a total incident photon flux of \( 5 \times 10^5 \) photons/pix as for all the simulations in this study. Also the reconstruction of the BT images has been performed with FBP and the use of Ram-Lak filter in sagittal and coronal direction.

![Image of ROIs from different planes](image)

**Fig. 7.10: ROIs from different planes, where the respective CaCO\(_3\) is in focus from the axial reconstruction**

In Fig.7.10 are illustrated some ROIs from the axial reconstruction at different depths, where each \( \mu \)C is in focus. Even though the incident beam was of 28keV, all 49 microcalcifications could be detected. As it can be seen in Fig. 7.10, \( \mu \)Cs are clearly visualized but the image appears to be blurred and to have lots of artifacts. These artifacts are due to the acquisition arc that was chosen. In BT images, artifacts are unavoidable since a 3D object is reconstructed from a limited number of projections over a limited angular range. The reconstruction algorithms generate out of plane artifacts, which appear as blurred versions of the in plane objects. For high density structures, like \( \mu \)Cs, these artifacts become stronger and appear progressively blurred in out of focus planes. Figure 7.11 shows the 7 planes from the sagittal reconstruction, where \( \mu \)Cs are in focus. This view allows clearly detecting the total of 49 \( \mu \)Cs.
As a result, a different acquisition arc compared with the conventional one can prove to be really useful. When the snail phantom was irradiated at 28keV with an arc -20°:2°:20°, μCs in the area of 13mm-33mm implant thickness could not be clearly detected. In addition, for the case of DE BT almost the total of features could be visualized, but the CNR was significantly low for high thicknesses of silicone gel. In contrast, the use of a discontinuous arc -80°:2°:-60° & 60°:2°:80° resulted in improved detection of the total of microcalcifications with sufficiently higher contrast.

7.5. Conclusion

In previous chapters the effect of silicone gel insertion in breast tissue and the inability of conventional imaging modalities to detect high contrast features in its neighborhood were shown. In this chapter images acquired with non-conventional breast imaging modalities were evaluated. Energies above the conventional area of breast imaging, DE subtraction mammography, DE subtraction BT and BT with a discontinues arc were tested.

The results from Dual Energy Subtraction Mammography showed the ability of this modality to visualize the total of μCs under a wide range of silicone gel thicknesses, up to 36mm for the step-wedge phantom and up to 49mm for the snail phantom. The tradeoff for this improvement in detectability is the reduction of CNR for small thicknesses of silicone gel. The combination of energies that resulted in the optimal detectability of high contrast features is the 28 keV-40 keV for both phantoms. The above combination
resulted in the highest mean value of CNR around 4, for thicknesses 10mm to 49mm. For the interval 1mm to 10mm, the optimal combination of energies was 20 keV-28 keV for the snail phantom and 20-80keV for the step-wedge phantom.

Dual Energy BT produced similar results with those of DE mammography. In contrast with mono-energy images, DE led to an improvement in detectability for high thicknesses of silicone gel; however, it caused a reduction in CNR for small thicknesses. The optimum results for DE BT were produced with the same set of energies as in DE mammography, i.e. 28keV-40keV. Another combination of energies that resulted in high mean CNR values for the case of the snail phantom was the 28keV-60keV. In this case, the pattern disappears while the total of 49 microcalcifications can be detected with a sufficient mean CNR (5 for small thicknesses and 2-3 for big thicknesses). Finally, there were no significant improvements of DE BT over DE mammography in the detection of small high contrast lesions. Undoubtedly the greatest advantage of DE BT lies in its potential to reconstruct both high and low contrast lesions in the presence of silicone gel.

By comparing the three imaging modalities, mammography, BT and DE, in terms of better visualization of high contrast features in the presence of silicone gel, DE dominates. Dual Energy with the appropriate combination of low and high energy and multiplier factor R was a modality that could detect the total of μCs in our experiments. Moreover seems to be an appropriate modality for breast imaging when an implant is inserted and both masses and microcalcifications need to be detected especially when DE BT is implemented.

Finally, a discontinues angular range -80°:2°:-60° & 60°:2°:80° was tested as a non-conventional BT acquisition. This alternative imaging modality improved the detection of lesions but caused too many artifacts to appear in the images as a result of the elongated arc.
8. INVESTIGATING PHASE CONTRAST AS A PROMISING TECHNIQUE FOR SILICONE EFFECTS LIMITATION

Ongoing research in innovative breast imaging modalities is growing in order to increase soft tissue contrast and spatial resolution (Russo et al 2016, Brenner et al 2007, Schleede et al 2014). Phase contrast imaging is such an emerging technique sensitive not only to attenuation but also to x-ray phase change arising at the boundaries of different refractive materials (Bravin et al 2013, Bliznakova et al 2012). X-ray photons interact with the tissues in the mammography energy range by one of the following main interaction mechanisms: the photoelectric effect, the Compton scattering and the coherent scattering. Photons that are coherently scattered deflect from their initial path at small angles. This kind of refraction of radiation occurs at the boundaries separating various media in an object of different x-ray refractive indices and lead to the phase contrast effect. Tissues composed of low Z-elements, such as breast masses, produce low absorption contrast but considerable phase contrast, resulting in significant edge enhancement. Schematic explanation of the mechanism of edge enhancement is depicted in Fig. 8.1.

![Diagram of edge enhancement](image)

**Fig. 8.1: Schematic presentation of the mechanism of edge enhancement due to x-rays refraction at the edge that separates two regions in an object.**

The x-ray beam passes through the edge separating two different regions of the object. In this particular example, it was assumed that the two regions have very similar absorption coefficients but different x-ray refraction. As shown in this figure, the x-rays refract from the edge and the corresponding beam deviation by the edge region produces a sharp white-black fringe pair on the detector that leads to improved edge visibility. This edge is very well visualized at sufficient distance between the object and the detector. Small and thin details that are not visible in absorption images become detectable as a result of this edge-enhancement effect. In order to visualize phase contrast effects, a sufficient degree of lateral coherence in illuminating radiation is required given by:
\[ d = \frac{\lambda \cdot R}{f} \quad \text{Equation 8.1} \]

where \( \lambda \) is the wavelength, \( R \) is the distance source to object and \( f \) is the size of the source. According to this relationship, successful phase contrast will be implemented with low energy photons, long distance source-object and very small source size. The amount of phase change in a tissue is related to its refractive index, which is given by:

\[ n = 1 - \delta + i\beta \quad \text{Equation 8.2} \]

where \( n \) is the complex refractive index of the tissue, \( \beta \) is the absorption index and \( \delta \) is the refractive index (Born and Wolf 1980). The coefficient \( \beta \) is responsible for the x-ray attenuation, while the coefficient \( \delta \) determines the phase change. The traditional absorption radiography depends only on the imaginary component \( \beta \), while the phase shift depends only on the real component \( \delta \). Figure 8.2 depicts the comparison between the real and imaginary parts, delta and beta respectively, of the complex refractive index of breast tissue.

![Graph showing the comparison between real and imaginary parts of complex refractive index](image)

**Fig. 8.2:** The real and imaginary parts, delta and beta respectively, of the complex refractive index of breast tissue.

It must be noted that \( \delta \) of tissues is greater than \( \beta \) by approximately two to three orders of magnitude for x-rays in the diagnostic energy range. Additionally, \( \delta \) diminishes less than \( \beta \) as the energy increases; thus the potential application of phase contrast imaging with higher energies (an unrealistic approach in case of conventional mammography) is possible with lower dose to the breast (Lewis 2004).

All these advantages of phase contrast imaging in contrast with conventional attenuation imaging may improve the visualization of lesions in case of a silicone gel presence in breasts. The fact that phase contrast enhances the edges of small details, which is important for the characterization of lesions in
mammography, and that their visibility is reduced in a silicone gel presence may be of a great importance in mammography of augmented breasts.

Phase contrast images may be acquired using one of the following four methods: In-Line Phase Contrast Imaging (PCI), known also as refraction enhanced imaging or free space propagation based imaging (Arfelli et al 2000, Castelli et al 2011), Diffraction Enhanced Imaging (Hasnah et al 2002, Pisano et al 2000, Fiedler et al 2004), X-ray Interferometry (Momose et al 1996) and Grating Interferometry (Pfeiffer et al 2006). Among these methods, the most popular is the In-Line PCI. This approach is the simplest phase contrast method and the one this thesis is focused on, since it does not require the use of any optics and grating elements between the sample and the detector in the imaging set-up, in order to perform wave splitting or any kind of image reconstruction. In addition, this technique has already been tested in clinical diagnosis.

In In-line PCI, a monoenergetic radiation is transmitted through the object. Parameters such as detector resolution, beam energy and object to detector distance (ODD) play an important role in the image quality. The final image is a recording of the intensity distribution at a certain distance behind an object and is created by interference of the scattered wave with the coherent incoming plane wave. This can be described by means of near field (Fresnel) diffraction theory. The intensity distribution at a distance $z$ from the object is given by:

$$I_z(r_{trans}) = I_i(r_{trans})|T(r_{trans})|^2 P_z(r_{trans})$$  \hspace{1cm} Equation 8.3

where $r_{trans}=(x,y)$, i.e. the transverse coordinates if we assume that $z$ is the x-ray propagation direction, $I_i(r_{trans})$ is the initial intensity of the wave, $T$ is the 2D transmittance function of the object and $P_z(r_{trans})$ is the Fresnel propagator. In the near field region (i.e. for $z<d^2/\lambda$ where $d$ is the size of the object and $\lambda$ is the wavelength), and assuming that the absorption is weak and slowly varying, the intensity measured at the detector can be written as follows:

$$I = I_0 T(1 - \frac{\lambda d}{2\pi} \nabla^2_{trans} \varphi)$$  \hspace{1cm} Equation 8.4

where $I$ is the intensity incident on the object and $\nabla^2_{trans} \varphi$ indicates the Laplasian in the $(x,y)$ plane of the phase shift introduced by the sample.

In order to visualize phase contrast effects, a sufficient degree of lateral coherence is required, dependent on the wavelength, the source to object distance (SOD) and the size of the source (Olivo et al 2009). That is currently well realized in synchrotron facilities (Lewis 2004). Synchrotrons radiation sources generate x-ray beams that are almost monochromatic, laminar (reduction on scattered radiation) and have a
sufficient spatial coherence to allow the use of free space propagation phase contrast imaging. However, as shown by Wilkins and associates (Wilkins et al 1996), it is largely insensitive to even broad polychromaticity in the near field diffraction regime, thus micro focus sources can be efficiently used.

### 8.1. Phase contrast simulation study

In order to investigate phase contrast mode as an imaging modality in case of silicone presence, simulations were performed. A software platform for phase contrast x-ray breast imaging research was used (Bliznakova et al 2015). The core of this platform method is of an analytical approach, based on Fresnel-Kirchhoff diffraction integral and paraxial approximation of the propagating x-ray waves (Peterzol A et al 2007). For the needs of this study, three similar rectangular slab phantoms were designed with dimensions of 15x15x40 mm³ composed of paraffin, as shown in Fig. 8.3.

![Fig. 8.3: Illustration of the software phantom used for phase contrast imaging examination.](image)

A cuboid with 5mm width, 10mm length and thickness varying between the three phantoms to 5 mm, 10 mm and 30 mm, composed of silicone gel was inserted in the center of the paraffin blocks. Three CaCO₃ spheres with radii of 0.1 mm, 0.2 mm and 0.3 mm representing μCs, as well as three water spheres with radii of 0.5 mm, 1mm and 1.5mm representing masses, were inserted under the silicone block. The construction of these phantoms allows investigations of both high (μCs) and low (masses) contrast lesions in the presence of varying thicknesses of silicone gel.
Two sets of simulations were performed for each of the three phantoms, resulting in attenuation and phase contrast images of 4000x4000 pix$^2$. In order to simulate an attenuation mode and a phase contrast mode, the IDD was set to 60 mm and 1 m respectively. The number of photons/pixel was set to 300 in order to result in an ESE of 2 mGy for all the experiments. The incident beam energy was 20 keV and the resolution of the image 200 pix/mm. Since the simulation was run with the analytical module, meaning that images were noise free, Poisson quantum noise was subsequently added to the original calculated images. A Poisson random generator was used, with a variance set equal to the number of photons that are incident on each detector pixel (Shinzato 2007). The resulting images are presented in Fig. 8.4. The first column shows the images acquired with an attenuation contrast mode and the second those acquired with phase contrast mode. The images are cropped in the borders of the silicone cube and presented after an appropriate windowing for better visualization of the six spheres.

In the attenuation image with the silicone block of 30 mm thickness, neither the masses nor the $\mu$Cs are visible. This is an outcome that coincides with previous results showing that mammographic details were not visible when silicone of this thickness was present. On the other hand, in the respective phase contrast image, two CaCO$_3$ spheres are slightly visible, signified in Fig. 8.4 with arrows. In the second row of Fig. 8.4, images with thinner silicone block of 10 mm acquired in an attenuation and phase contrast mode are illustrated. As it can be seen, in the attenuation image the smaller water and CaCO$_3$ spheres are still not visible, while the rest starts to appear. But in the phase contrast image all the $\mu$Cs are visible, even the smallest one, whereas the two masses start to be distinguished. Finally, in the third row with a 5 mm silicone block, all $\mu$Cs are detectable in both attenuation and phase contrast images, whereas the masses are still difficult to be detected in attenuation mode. Even though in both modes all the CaCO$_3$ spheres are detectable, there is a significant difference in their borders between the two modalities. In the attenuation image the smaller spheres are characterized with reduced visibility and unclear shape, while in the phase contrast image their edges are enhanced.
Fig. 8.4: Images acquired in an attenuation (first column) and phase contrast (second column) mode with different silicone thicknesses.
Figure 8.5 provides a better understanding of this result, illustrating \( \mu \)Cs spheres of 0.1 mm and 0.2 mm radii in both modalities.

![Figure 8.5: Regions of interest for the CaCO3 sphere (first column) and the water sphere (second column) for both an attenuation and a phase contrast mode.](image)

From all the above results it can be concluded that phase contrast imaging can improve the visualization of high and low contrast features at silicone presence and enhance their visibility compared with conventional breast imaging. Moreover, the fact that phase contrast imaging results in an edge enhancement may improve the identification of breast lesions. In the next chapter, phase contrast imaging techniques will be investigated in detail as a promising technique for the enhancement of breast structures.
9. IMAGE QUALITY EVALUATION OF DIGITAL MAMMOGRAPHY TECHNIQUES WITH PHASE CONTRAST SYNCHROTRON RADIATION

9.1. Introduction

Cancer incidence and mortality statistics show that breast cancer is one of the most common types of cancer worldwide, accounting for 3 in 20 (15%) cancer deaths in females (Cancer Research UK). Mammography is the standard modality of breast screening and diagnosing. Although mammographic screening techniques have increased the image quality, the rate of missed lesions and false positives remain crucial (Pissano et al 2014, O'Conneli et al 2010).

Despite technical improvements in x-ray sources and digital detectors, mammography is far from being perfect (Gong et al 2006). In principle, mammography projects the compressed breast structure onto a two dimensional plane, based on x-ray tissue absorption. Anatomical structures can overlap resulting in poor visualization of useful diagnostic information and insufficient soft tissue contrast.

Early experimental studies on surgically removed breast specimens that contained tumor nodes showed that mammograms obtained with synchrotron radiation have higher contrast and better resolution in comparison with traditional mammograms and demonstrated better detail in all cases studied at a dose equal to the dose used in standard mammography (Burattini et al 1995). Besides the experimental work with simple and thin objects, a number of studies investigated this technique on breast specimens with microcalcifications (Ingal et al 1998, Arfelli et al 2000, Kiss et al 2004, Pagot et al 2005, Imamura et al 2008), lobular or ductal carcinoma (Fiedler et al 2005). All studies reported that PCI enhances the contours of different tissues, as the contrast is enhanced and spatial resolution is increased with better definition of glandular component and improved visibility of calcifications.

In the previous chapter the advantages of phase contrast imaging versus conventional imaging in the case of silicone presence were shown through simulations. This chapter focuses on the potentials of combining
the advantages of BT and phase contrast for breast imaging. Specifically, the in depth information without overlapping anatomy that BT provides is combined with the edge enhancement effect resulting in a superior contrast of malignancies arising as a phase contrast effect. In order to do so, a phantom was constructed of paraffin wax mimicking a compressed breast geometry, where nylon fibers, spheres and CaCO$_3$ powder were embedded. Projection images of the phantom acquired at two object to detector distances of 50cm and 150cm in a mammographic and a BT mode were compared visually and quantitatively. All the experiments were performed at the ELETTRA Synchrotron Light Laboratory, Trieste, Italy, using synchrotron radiation of 20 keV. Two acquisition arcs were used for the BT mode, a narrow one of 15° and a wide one of 44°. Different figures of merits were used in order to compare the images acquired with different modalities.

9.2. Hardware Phantom

For the needs of this study a phantom was constructed in our facilities based on a former study about the materials and the geometry of a breast-like phantom. A wide range of different compounds were investigated in order to find the optimum materials to represent fibroglandular tissue, adipose tissue, skin, lymph, microcalcifications and breast masses for an energy range between 18keV and 40keV. A comparison of the $\delta$ values between these breast tissues and the mimicking materials has been calculated according to the equation:

$$\delta_{\text{Diff}}=\left(\frac{|\delta_{\text{BT}}-\delta_{\text{M}}|}{\delta_{\text{BT}}}\right)\times 100$$

where $\delta_{\text{BT}}$ represents the refractive index of the breast tissue and $\delta_{\text{M}}$ the refractive index of the mimicking material (Vedantham and Karellas 2013). The phantom constructed consists of an outer cylinder of 5 cm radius and 2.8 cm thickness made of paraffin wax (C$_{25}$H$_{52}$, density: 0.93 g/cm$^3$, $\delta_{(20\text{keV})}=5.53 \times 10^{-7}$), inside which are embedded three different types of abnormalities. Nylon spheres with sizes 3/16 inch, 1/8 inch and 3/32 inch representing masses, nylon fibers with sizes 0.9 mm, 0.7 mm and 0.5 mm, and CaCO$_3$ powder representing microcalcifications ($\mu$Cs). Figure 9.1 presents a photograph of the phantom at its final stage and a mammographic image acquired at 20 keV with synchrotron radiation. In order to avoid the presence of air bubbles in the base material, the paraffin wax was melted at 70° C under vacuum. During the solidification process four layers of the abovementioned abnormalities were embedded slowly.
The aim of this arrangement was to produce areas with overlapping structures in order to facilitate a BT investigation.

Fig. 9.1: Illustration of the phantom at his final stage (on the left) and a mammographic image acquired at 20 keV (on the right).

9.3. Image acquisition and reconstruction process

All the experiments presented in this chapter took place at the ELETTRA Synchrotron Trieste, at the SYRMEP beamline, especially designed for research in medical diagnostic radiology.

Fig. 9.2: Schematic drawing of the beamline optics.
The optics are based on a double-crystal Si(111) monochromator working in the energy area 18-35 keV. The x-ray beam provided at 20 m from the source is a laminar section with maximum area 120 x 4 mm². In Fig. 9.2 is provided a schematic representation of the beamline optics. The crystals assembly is equipped with high precision motion stages to set the Bragg angle and to perform the final angular alignments of the second crystal with respect to the first. The fixed exit of the beam, independent from the selected energy, is obtained by translating the second crystal along a linear guide which makes an angle of about 3° with respect to the beam direction. The horizontal acceptance covered by the light-port of the front-end is 7 mrad. Prior to the monochromator, the first element after the front-end is a 2 mm beryllium (Be) window, which divides the ultra-high vacuum of the ring from the beamline and eliminates the low frequency components of the radiation. This window is followed by a slit system that allows selecting the beam size and shape. A second 0.5 mm Be window is located after the monochromator and separates the vacuum components of the beamline from the rest of the set-up.

Fig. 9.3: Photograph taken during the experiments at the ELETTRA synchrotron facilities.

Then, a second system of slits is used to shape the beam and stop the radiation scattered by the previous elements. Finally, a system of two identical ionization chambers measures the intensity of the radiation while safety shutters are used to control the exposition during the experiments. The beam then enters the
experimental room (Fig.9.3). The energy of the x-ray beam was 20 keV for the total of the experiments of this study. The phantom was placed on a metal plate moving vertically at each scan and able to rotate 360° fixed at 22.4 m from the source.

The detector was a Teledyne DALSA TDI CCD camera (DM-20-08K10-00-R, 16 bits, 54µm pixel-obtained after 4x4 binning mode, 22 cm imaging area). The experiments were performed with two object to detector distances (ODD) of 50 cm and 150 cm, resulting in a magnification of 1.022 and 1.066 respectively. With both arrangements, a free space propagation phase contrast imaging set-up was produced. The first set-up with ODD 50cm resulted in images where the phase contrast contribution started to appear but cannot be fully seen, whereas the second one with ODD 150cm resulted in images with strong edge enhancement. Two different modalities were used to produce images of 4400x2200 pixels. The first one was a mammography mode, where 2D images were acquired. The second one was a BT mode at two different acquisition arcs: one with an acquisition arc of 14° and 1° increment resulting in 15 projections (Hologic performance) and another one with an acquisition arc of 44° and 2° increment resulting in 23 projections (Daskalaki et al 2016). In our experiments two mean glandular dose (MGD) ranges were chosen. The first one was 2.1-2.6 mGy, according to the European guidelines for quality assurance in breast cancer screening and diagnosing; the second one mentioned in the text as Max Dose, was sufficiently increased at 34 mGy. The reason for such an increase was the investigation of phase contrast effects relative to dose values. For the BT mode, the MGD was equally distributed in the projections. The incident air kerma was calculated in advance for all the cases in order to acquire an MGD in the above mentioned dose ranges. For the calculations, the following equation is used:

\[ MGD = K \times g \times c \times s \]  

Equation 9.1

where g, c and s are conversion parameters depending on the breast thickness, the glandularity and the incident spectra. In our calculations, those parameters were taken from Dance (Dance et al 2000, Dance et al 2011). The incident air kerma (K) is measured at the upper surface of the breast. In our study, where
the base material of the phantom is paraffin wax, the equivalent breast thickness was calculated and found to be 2cm.

Before reconstruction, a flat-field correction of the BT images was performed. Additional processing needed in order to correct the small vertical movements of the phantom during the BT acquisition arcs, was done using MATLAB 2013. For the reconstruction process an in-house platform was used (Kamarianakis et al. 2014) performing filtered back-projection. All the images were pre-filtered with the use of Ram-Lak filter. A pseudo-3D representation of the phantom was produced from the reconstructed axial planes at every 0.1 mm of the total volume.

**9.4. Evaluation metrics**

In order to evaluate the image quality and the detection of breast abnormalities, two evaluation metrics were introduced. Those metrics were based on the Contrast (C) and SNR widely used for image quality assessments in conventional (attenuation based) imaging as they were presented in Chapter 3.4.4. For phase contrast imaging, where the edge enhancement needs to be accounted in the calculations, those metrics are redefined as $C_{PC}$ and $SNR_{edge}$ respectively, based on Equations 9.2 and 9.3 (Diemoz et al. 2012).

$$C_{PC} = \frac{I_{max} - I_{min}}{I_{back}}$$  \hspace{1cm} \text{Equation 9.2}

$$SNR_{edge} = \frac{I_{max} - I_{min}}{\sqrt{2} \times \sigma_{back}}$$  \hspace{1cm} \text{Equation 9.3}

where $I_{max}$ and $I_{min}$ are the maximum and minimum values of the intensity profiles (Fig. 9.4). $I_{back}$ and $\sigma_{back}$ are the average intensity value and the standard deviation of the background ROI respectively, both calculated over the rectangle area.

For the calculation of those metrics, regions of interest (ROIs) were selected, as it can be seen in Fig. 1. Those ROIs were actually horizontal line profiles for masses (M1,M2,M3) and $\mu$Cs (C1,C2,C3) passing
almost at the center of the feature avoiding manufactured defects, whereas for fibers they were squares (Prof1-3). For each of the evaluated features a Background rectangle ROI of 5304 pix$^2$ was selected in an area nearby without any abnormalities insertion. Out of the resulting 12 Background ROIs, one is presented in Fig. 9.4.

![Fig. 9.4: Projection of the phantom where the ROIs of the findings under investigation are illustrated.](image)

In order to examine the small variations of dose and allow the comparison of SNR$_{edge}$ at each experiment, a figure of merit (FOM) was introduced, where the SNR$_{edge}$ was normalized to the square root of the mean glandular dose (MGD) according to Equation (9.4).

$$FOM = \frac{SNR_{edge}}{\sqrt{MGD}}$$  \hspace{1cm} \text{Equation 9.4}
9.5. Results

Regions of interest as well as evaluation metrics were used for visual and quantitative assessment of the images acquired. Figure 9.5 presents ROIs from 2D images of the phantom acquired with ODD 50 cm and 150 cm, where μCs, fibers and masses appear almost separately.

Table 9.1: Values of FOM for the ROIs presented in Fig. 9.4 at eight different imaging modalities.

<table>
<thead>
<tr>
<th>FOM</th>
<th>2D</th>
<th>ODD=50cm</th>
<th>ODD=150cm</th>
<th>ODD=150cm MAX Dose</th>
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<td></td>
<td></td>
<td>Arc 15°</td>
<td>Arc 44°</td>
<td>Arc 15°</td>
</tr>
<tr>
<td>M1</td>
<td>2.1</td>
<td>2.4</td>
<td>2.3</td>
<td>4.4</td>
</tr>
<tr>
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<td>1.5</td>
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</tr>
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<td>8.1</td>
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<td>7.7</td>
<td>7.8</td>
<td>11.9</td>
</tr>
<tr>
<td>C3</td>
<td>8.3</td>
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Table 9.1 shows the FOM values calculated over the ROIs presented in Fig. 9.4 for the eight different imaging modalities investigated in this study. M1,M2,M3 and C1,C2,C3 correspond to ROIs of the masses and μCs from the smallest to the biggest one respectively, whereas the six "Prof" correspond to ROIs of fibers from the biggest one "Prof1a-b" to the smallest one "Prof3a-b" (Fig. 9.4).

As it can be seen in Fig. 9.5, 2D images acquired with ODD 150 cm show a strong edge enhancement of the abnormalities that is barely seen in images acquired with ODD 50 cm.
Fig. 9.5: Regions of interest for μCs, fibers and masses from the 2D images acquired with an ODD 50 cm (first column) and ODD 150 cm (second column).

Actually, the edges in the latter case appear blurred, whereas in increased detector distance they appear sharp and with an improved contrast. Moreover, in the case of ODD 150 cm, the fine internal structures of the masses and μCs are revealed. In addition, according to Table 9.1, for 2D images FOM values are higher at increased ODD for all the findings. This improvement of FOM is significantly high for masses, since it reaches 43% on average, whereas for μCs and fibers it is 21%. A diagram of the averaged FOM
for each type of abnormality at each imaging modality is presented in Fig. 9.6 that will aid to compare the images acquired in terms of equal doses (MGD).

![Fig. 9.6: The averaged FOM for each type of abnormality at the eight different imaging modalities investigated.](image)

In Figures 9.7, 9.8 and 9.9, phase contrast (ODD=150cm) BT planes with different cluster of μCs, fibers and masses appearing in focus are presented in columns as we slice through the reconstructed volume. The first, second and third row of those figures present slices from the BT reconstructed volume acquired with an arc of 15°, an arc of 44° and an arc of 15° but with sufficiently higher dose, respectively. The arrows on the figures signify the structures appearing in focus on their plane. Table 9.2 shows the contrast measurements of masses and μCs for the different BT acquisitions. As well as for the 2D case, visual assessment of the BT volumes and quantitative results of FOM and contrast values showed that higher ODD results in improved phase contrast images and superior visualization of lesions. Moreover, BT slices acquired with the narrow arc of 15° exhibit higher FOM and contrast values for the total of the findings embedded in the phantom.
BT slices resulted in an improved detectability of small structures that were barely seen in 2D images. The detailed examination and the differentiation of the abnormalities according to their geometries led to the results presented below. The great increase of dose resulted in far better edge enhancement of lesions. On the other hand, by maximizing the dose, the contrast of features degrades and the FOM reduces dramatically, even though the enhancement of the edges appears stronger. Although the in plane structures appear blurred as we slice through the volume (common effect of BT), they are losing their edge enhancement.
Fig. 9.7: Phase contrast (ODD=150cm) BT planes with different clusters of μCs appearing in focus presented in columns as we slice through the reconstructed volume.
Fig. 9.8: Phase contrast (ODD=150cm) BT planes with different fibres appearing in focus presented in columns as we slice through the reconstructed volume.
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Fig. 9.9: Phase contrast (ODD=150cm) BT planes with different masses appearing in focus presented in columns as we slice through the reconstructed volume.
9.6. Discussion

Visual assessments of the resulting images as well as FOM and Contrast measurements of the main findings embedded in the phantom showed that the increase of ODD results in improved image quality. In addition, the edge enhancement of the main findings becomes stronger and their appearance is sharper. These results may be critical in the detection of small structures. For instance, the cluster of μCs seen in Fig. 9.5 inside the black circle can be clearly depicted only in images acquired with high ODD, whereas in the rest it cannot be detected among the background noise. Moreover the improvement of FOM was higher for masses, that may contribute in their detectability compared with attenuation based imaging where is difficult to be identified especially in dense breasts.

By comparing the 2D images with the respective BT slices, the latter resulted in an improved visualization of lesions. The biggest advantage of BT over mammography lies in the decrease of structural noise, a problem caused by overlying structures (Dobbins and Godfrey 2003). In Fig. 9.5 black squares signify the regions with superposition of structures in 2D images. Area A in Fig. 9.5 illustrates a superposition of μCs that in Fig. 9.7 is resolved and illustrated at two separate BT planes (first and last row). Similarly, area B in Fig. 9.5 illustrates an overlap of two fibers that was impossible to be distinguished in the 2D image. In the second and third row of Fig. 9.8, however, they are resolved and identified at different planes. The same stands for area C of Fig. 9.5 with the two masses (4.8 mm) overlapping, which are illustrated separately in the second and last row of Fig. 9.9. Moreover, according to Table 9.1 and Fig. 9.6, BT images resulted in higher FOM values compared with 2D images, especially for μCs.

A comparison of the two acquisition arcs chosen for phase contrast BT investigation showed that the narrow one (15°) resulted in higher values of FOM and contrast. The lesions appear sharper and the contrast of their edges is more intense. This is an important result for breast imaging, since the enhanced contrast of low contrast features as well as the clear visualization of μCs needs to be improved. Moreover, the 15° arc BT resulted in out of plane artifacts less elongated as a result of the smaller arc. On the other hand, the contrast of those out of plane artifacts is higher in 15° arc BT, it may therefore be misinterpreted as in plane structures. This drawback is more intense for high contrast features, as it can be seen in Fig. 9.7. In addition, the case of the BT with the arc of 44° showed better in depth resolution, thus resulting in an improved in depth localization of features that might be helpful in biopsy guidance and breast operation planning. Some examples of this improved in depth resolution can be seen in Fig. 9.8 (second column), where the fibers are greatly resolved from each other as we slice through the 44° arc BT volume.
The artifact spread function (ASF) was used to quantify the "out of plane" artifacts observed in planes outside the plane of focus (Zhang et al 2006, Lu et al 2011) according to Equation 9.5:

$$ASF(z) = \frac{CNR(z)}{CNR(z_0)}$$  \hspace{1cm} \text{Equation 9.5}

$$CNR = \frac{|I_f-I_{back}|}{\sqrt{\sigma_f^2-\sigma_{back}^2}}$$  \hspace{1cm} \text{Equation 9.6}

where the CNR was calculated according to Equation 9.6 and $z_0$ is the location of the in focus plane and $z$ is the location of the off focus plane, respectively. Figure 9.10 illustrates the ASF calculated according to Equation 9.5, from the intensity of in-focus and off-focus ROI positioned inside calcification C3 (Fig. 9.4) for the 6 different cases of BT used in this study. The superiority of BT in diminishing out of focus artifacts for the case of arc 44° compared with the one of 15°, was well confirmed by the ASF values presented in Fig. 9.10.

Fig. 9.10: A diagram of the ASF calculated for calcification C3 at the six BT cases.

The values of contrast presented in Table 9.2 for ODD 150 cm dramatically reduced when the MGD was increased (MaxD) for both masses and μCs. Moreover, the SNR$_{edge}$ of the main findings increased with
higher values of MGD and so did the enhancement of their borders as it was expected, but the corresponding FOM reduced (Table 9.1). The above result comes in agreement with (Gonzalez et al 2016) stating that the best image quality was not obtained using the highest dose. In addition, as the MGD increased, so did the contrast of the out of plane structures in BT slices which may be proven disturbing. Both results indicate that phase contrast imaging may be used in clinical practice producing high quality images with low doses delivered to patient, and that there is a limit for patient radiologic protection optimization, but further investigation on dose analysis is needed.

9.7. Conclusion

This study evaluated the potentials of combining the advantages of BT and phase contrast for breast imaging. In order to do so, a phantom was constructed made by paraffin wax, whereas three different details, spheres, fibers and CaCO$_3$ powder of various dimensions, were embedded. Breast tomosynthesis and 2D images of the phantom in a phase contrast mode at two ODD distances were acquired in order to compare them visually and quantitatively. Evaluation metrics were redefined in order to account the edge enhancement of the details under investigation. All the images were acquired with the use of synchrotron radiation at 20 keV and an MGD in a conventional breast imaging area. Moreover, the effect of the dose was investigated by increasing it out of medical dose limits.

Phase contrast images resulted in a superior contrast with the increase of ODD, for both μCs and masses. In addition, the increase of ODD resulted in a strong edge enhancement of all the features proved by the comparison of FOM values. On the other hand, BT images with increased ODD, not only achieve to visualize μCs that either slightly appear or are not visible in 2D images, but they also eliminate the overlapping effect that dominates the 2D projection images and make possible the in depth localization of structures. Finally, BT images demonstrate higher values of all the evaluation metrics compared with mammographic images. All the above results conclude that BT phase contrast imaging is a promising technique that can be proved important for the detection of small details in breast screening and diagnosing.

A comparison of the two acquisition arcs for BT showed that a narrow arc (15°) resulted in better FOM and contrast values for the main findings but in inferior in depth localization and elimination of out of focus structures compared with the wide arc (44°). This result is in agreement with (Malliori et al 2012) stating that low and high-contrast features are better visualized at extended arc lengths. This indicates that
further investigation for the optimal acquisition arc in the case of phase contrast BT is necessary and may trigger post image processing techniques. In addition, dose analysis showed that by increasing the delivered MGD, the SNR_{edge} of all the features was increased, whereas the contrast was reduced. Finally, FOM values revealed that lower MGD values resulted in superior image quality, an important finding for dose reduction in the case of phase contrast BT.
10. CONCLUSION

This study evaluated the quality of images acquired with conventional breast imaging techniques in the presence of silicone gel. The detectability of both high contrast features, like μCs, and low contrast features, like breast masses, is examined on simulated mammography and BT images in the presence of silicone gel of different thicknesses at beam energies in the interval of 20-30 keV. Three complicated software phantoms with silicone gel insertions were designed for x-ray imaging studies. These phantoms were designed with escalating geometries in order to investigate the effects of different silicone thicknesses in a single imaging phantom. Since no such phantoms are commercially available, their future realization will be an essential tool for image quality assessment of mammograms, optimization of imaging protocols including breast tomosynthesis, and dosimetry, in the case of implant presence in the breast.

The results of this experiment clearly show how a silicone insertion affects the image quality of a mammogram and how those images lack contrast depending on the thickness of silicone gel. Lesions placed under silicone gel material with thickness higher than 26 mm could not be visualized either with mammography or BT, revealing the problems arising from the implant insertion in breast imaging. Simulations showed a dramatic reduction on the CNR values for the detection of μCs as a result of the silicone presence. This reduction is in the order of 45% for a silicone overlap of 2 mm and increases with the increase of silicone thickness. Conventional mammographic images resulted in superior CNR values compared to those calculated for tomosynthesis. This is especially valid for μCs lying under thin implants imaged with low beam energies. Mammography also resulted in higher limits of silicone gel thickness for μCs detection compared with BT.

At low energies (20-22 keV), CNR measurements showed high dependence on implant thickness, while the response of CNR rapidly decreases as the implant thickness increases. This dependence of CNR on the implant thickness becomes less steep with the increase of energy. Alterations of energy in the interval of 20 keV to 30 keV led to improved detectability of μCs overlapped by silicone, while the opposite stands for breast masses. Although BT images resulted in inferior image quality in terms of CNR, they demonstrated an advantage in visualizing a larger breast area and small low contrast lesions because of non-overlaps. Furthermore, BT results showed interesting behavior for certain heights in the snail and the step-wedge phantoms, which is attributed to the viewing angles of the acquisition arc. This indicates that the acquisition arc selected in an application where breast screening with silicone presence is needed, is critical. In addition, in some cases the μCs lying under high thicknesses of silicone were detected in BT images, indicating that BT presents some advantages compared to mammography.
Further simulations with the use of a breast phantom with complex texture under compression confirmed the clear disadvantage in the detectability of lesions and the lack of CNR due to silicone placement in the heterogeneous breast tissue as well. A comparison between the results with the complex phantom and the results with homogeneous background phantoms showed a further reduction in CNR values for lesions in the first case. This reduction is a result of the complexity of the base material where the lesions were inserted. Moreover, this reduction might not be so critical for the detection of μCs but for the mass, results are disappointing. The complexity and the density of the background in correlation with the scatter of silicone made it difficult to detect the mass even if its position was known a priori. As a result, it is shown that the complexity as well as the density of the breast texture will worsen the visualization of lesions in the case of augmented breasts.

Alternative imaging modalities were investigated through simulations in order to investigate whether image quality could be improved. Among these modalities, Dual Energy mammography and Dual Energy tomosynthesis were tested. The results from Dual Energy subtraction mammography showed the ability of this modality to visualize the total of μCs under a wide range of silicone gel thicknesses, up to 36 mm for the step-wedge phantom and up to 49 mm for the snail phantom. The tradeoff for this improvement in detectability is the reduction of CNR for small thicknesses of silicone gel. The combination of energies that resulted in the optimal detectability of high contrast features is the 28 keV-40 keV for both phantoms. The above combination resulted in the highest mean value of CNR around 4, for thicknesses from 10mm to 49mm. For the interval 1mm to 10mm, the optimal combination of energies was 20 keV-28 keV for the snail phantom and 20-80keV for the step-wedge phantom.

Dual Energy BT produced similar results with those of DE mammography. In contrast with mono-energy images, DE led to an improvement in detectability for high thicknesses of silicone gel; however, it caused a reduction in CNR for small thicknesses. The optimum results for DE BT were produced with the same set of energies as in DE mammography, i.e. 28 keV-40 keV. Another combination of energies that resulted in high mean CNR values for the case of the snail phantom was the 28 keV-60 keV. In this case, the total of 49 microcalcifications was detected with a sufficient mean CNR (5 for small thicknesses and 2-3 for big thicknesses). Finally, there were no significant improvements of DE BT over DE mammography in the detection of small high contrast lesions. Undoubtedly the greatest advantage of DE BT lies in its potential to reconstruct both high and low contrast lesions in the presence of silicone gel.

By comparing the three imaging modalities, mammography, BT and DE, in terms of better visualization of high contrast features in the presence of silicone gel, DE dominates. Dual Energy with the appropriate combination of low and high energy and multiplier factor R was the modality that could detect the total of
μCs in our experiments. Moreover, it seems to be an appropriate modality for breast imaging when an implant is inserted and both masses and microcalcifications need to be detected, especially when DE BT is implemented.

Results from conventional BT with the use of escalating phantoms showed that the acquisition geometry in correlation with the phantom's geometry may visualize lesions covered by high thicknesses of silicone gel that in other cases could not be imaged. For this reason, another imaging modality technique that was investigated was a BT with a discontinuous acquisition arc -80°:2°:-60° & 60°:2°:80°. This alternative imaging modality improved the detection of lesions with sufficiently high CNR compared with the rest of the imaging modalities studied. On the other hand, it caused too many artifacts to appear in the images as a result of the elongated arc. These artifacts may be diminished with the use of post filtering or masking.

The last imaging modality which was examined as an alternative method for imaging augmented breast was phase contrast. Simulations concluded that phase contrast imaging can improve the visualization of high and low contrast features at silicone presence and enhance their visibility compared with conventional breast imaging. Moreover, the fact that phase contrast imaging resulted in an edge enhancement may improve the identification of lesions.

The experimental study performed at Elettra Synchrotron facilities, in order to evaluate the image quality of breast lesions using phase contrast imaging, indicated a strong enhancement of breast structures. For the needs of this study a phantom made of paraffin wax was constructed, whereas three different details, spheres, fibres and CaCO$_3$ powder of various dimensions, were embedded. Breast tomosynthesis and 2D images of the phantom in a phase contrast mode at two ODD distances were acquired in order to compare them visually and quantitatively. Evaluation metrics were redefined in order to calculate the edge enhancement of the details under investigation. All the images were acquired with the use of synchrotron radiation at 20 keV and an MGD in a conventional breast imaging area. Moreover, the effect of the dose was investigated by increasing it out of medical dose limits.

Phase contrast images resulted in a superior contrast with the increase of ODD, for both μCs and masses. In addition, the increase of ODD resulted in a strong edge enhancement of all the features, which was proved by the comparison of FOM values. On the other hand, BT images with increased ODD not only achieve to visualize μCs that either slightly appear or are not visible in 2D images, but they also eliminate the overlapping effect that dominates the 2D projection images and make possible the in depth localization of structures. Finally, BT images demonstrate higher values of all the evaluation metrics compared with mammographic images. All the above results conclude that BT phase contrast imaging is a
promising technique that can be proved important for the detection of small details in breast screening and diagnosing.

A comparison of the two acquisition arcs for BT showed that a narrow arc (15°) resulted in better FOM and contrast values for the main findings but in inferior in depth localization and elimination of out of focus structures compared with the wide arc (44°). This indicates that further investigation for the optimal acquisition arc in the case of phase contrast BT is necessary and may trigger post image processing techniques. In addition, dose analysis showed that by increasing the delivered MGD, the $\text{SNR}_{\text{edge}}$ of all the features increased, whereas the contrast reduced. Finally, FOM values revealed that lower MGD values resulted in superior image quality, an important finding for dose reduction in the case of phase contrast BT.

The results of the experimental work for BT phase contrast imaging at high distances indicated an improved image quality of breast lesions, which can be proven important in the case of imaging breasts with silicone gel insertion. In the future, an experiment with a breast phantom and a silicone insertion would assist to confirm the above statement and verify the simulation results presented in this study.
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