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Evaluation of the Effect of Breast Density on Tumour Detection using X-ray Mammography: A Geant4 Simulation Study

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ABSTRACT

Screening mammography claimed to be associated with an increased risk of breast cancer. To justify screening, only a small increase in diagnostic performance can contribute to the preservation of an overwhelming number of women. Thus, optimal image quality must be taken into account on screening program. Breast density correlated with the accuracy of X-ray mammography and used as an indicator to convey information about the difficulty of tumour detections. The risk-benefit investigation should, focus on the performance of the examination. A well-known simulator, the modern Geant4 code, was used to model all the X-ray tube components of a molybdenum (Mo_{42}) target / filter (Mo_{42}/Mo_{42}) . In addition,

heterogeneous breast phantoms with thickness of 4 cm and different ranges of glandular tissues starting from 5% glandularity up to 100% simulated. At first all, the simulations validated by obtaining the energy spectra of molybdenumk_{α} = 17.48 keV, k_{β} = 19.61 keV

with a thickness of 50µm in the interval between 10 to 28 keV. In all cases, a tumour of 5 mm in diameter implanted within prevalence regions of glandular tissue. All obtained data evaluated to find a relationship between small-size tumour detection at different concentration of glandular tissue as well as the possible improvement of mammography image quality. The system performance assessed in terms of the Contrast-to-Noise Ratio (CNR) in multiple regions within the breast phantom. Results, suggest higher CNR values at lower mammary gland tissue ratio. Breast model was low with concentration >50%. Particularly, in position 1 the value was 0.026 but 0.26 at position 2. A high concentration at <50% (position 1=0.051 and position 2=0.348). CNR confirmed a strong negative association with all different mammary gland density concentrations. CNR predicted and used for measuring the diagnostic reliability of mammography tumour detection. Results suggested that the concentration of the glandular tissue i.e. breast density is the appropriate measure to describe the risks to which the radiation is exposed. In this case, an accurate estimate of the glandular tissue fraction must be known.

Keywords: Mammography Tube Modelling; X-ray spectra; Breast Cancer; Geant4; Glandular Fraction; Breast Density

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INTRODUCTION

Breast cancer is the most common cancer among women worldwide. It has a high mortality rate but this can be reduced if the lesion detected early [1]. Early detection of breast cancer at an early stage of tumour growth is more effective in treatment and consequently reduces the mortality rate by approximately 25% [2]. X-ray Mammography is one of the best technique for early detection of breast abnormalities with minimal radiation dose [3]. However, such imaging technique suffers from a reduction in sensitivity particularly when imaging women with dense breast. This what constitutes a risk of contracting carcinogenesis associated with the absorption of X-ray in the mammary gland, and considered the most radiosensitive tissue at risk [4]. As the ratio of fibroglandular, (dense) tissue increases in the breast tissue the sensitivity of mammography decreases. Thus, the radiopaque fibroglandular tissue can mask cancers. In other words, women with higher fibroglandular tissue were more likely to develop an increased in breast cancer by 4-6 times [5]. Several studies [6, 7] on density classification system predict the incidence of breast cancer for internalized risks associated with heterogeneously dense and extremely dense breast tissue. Of those, some provide an indication of the likelihood that cancers obscured. This may further emphasize that even the smallest regions of dense tissue in otherwise fatty breasts could obscure small lesions [4]. The difficulty in detecting tumour tissue is due to the similarity in linear attenuation coefficients between tumour tissue and normal mammary tissue. Thus, make it difficult to distinguish tumour tissue from mammary gland, especially at a high concentration of the mammary gland [8]. This requires studying image quality to improve lesions detectability at low-contrast. Thus, objective image quality metric should incorporate the measurement of the contrast-to-noise ratio (CNR) in selected Regions of Interest (ROIs) indicating the defect/tumour region [9].

GEANT4 Code released on the 29 of May 2020 used to model the X-ray mammography tube with target / filter. A (Mo_{42}/Mo_{42}) and anode angle of 17 degrees with beryllium (Be_4) window of 0.3 mm were included. The added filtration of molybdenum (Mo_{42}) was 30µm. The validity of the simulated data checked by comparing the calculated spectra with the one published in reference ten [10]. The code designed and optimized to operate at low energy of 28 keV. Geometric modeling of the proposed design based on Cone-beam breast X-ray techniques by placing the breast in a plane on a flat panel. Amorphous selenium (a-Se) detector with a pixel size of 0.5 mm modelled. The simulated data drawn in the Matlab program. All the obtained images of various glandular fraction distributions evaluated in term of the image quality. The images illustrate the distribution of the detected photons representing tumour contrast within

the breast. The effect on image quality clearly identified by investigating the Contrast-to-Noise Ratio (CNR).

MATERIALS AND METHOD

Geant4 simulation

As technology advances, Monte Carlo simulations increasingly used in several applications, such as astrophysics, radiation protection, and medical imaging [11]. This is due to its build in functions and tools that simulate all the physical processes including the interactions of electrons and photons with matter. It simulates all the energy down to the energy cut off 1 keV. This make it suitable for simulating the X-ray mammography systems. This worked is based on the GEANT4 toolkit which is a free software package written in C ++ programming. It originally developed at CERN [12], but now provides a complete set of software components for all aspects of the simulation process. Including full descriptions of the chosen geometry, the involved materials and the particles of interests. All the physical processes and the generations of all event data are possible. It also offer a full 3D visualization of the geometry and particle trajectories. The toolkit contains a set of physical models describing how particles interact with different materials [13]. This serves as a core layer and may be a good option for future simulations of experimental conditions in X-ray mammography tumour detection [14]. Low energy physical models included in the extensions of GEANT4 toolkit employed in this work. The production threshold ("range cut") fixed for the secondary particles expressed in terms of the distance travelled by the particles in the medium. Then converted by GEANT4 in terms of energy. For instance, the range cuts off $3000 \ \mu m$ for photons and $500 \ \mu m$ for electrons at different proportions of glandular breast tissue. All energies involved in these cases Rayleigh, Photoelectric effect, Compton scattering and Bremsstrahlung were simulated [15]. Characteristic and Bremsstrahlung radiation produced and counted. Rayleigh, Compton, Photoelectric interactions also included in the transport of photons through the simulation process. The spectral analysis of photon beam performed by simulating amorphous selenium (a-Se) detector positioned in the radiation field. Molybdenum filter placed in the radiation field to produce the desired attenuation [16].

Beam Geometry

One hundred millions primary particles created from the electrons source. Defined as a twodimensional 2D) of ~ $[100 \times 100]$ nm² planar source with a uniform distribution in the X-Y plane. It was placed at a position of 10 mm from the anode. Noting the Z-axis, projecting an actual focal spot size of ~ $[100 \times 100]$ nm² on the anode, as viewed on the image (X-Y) plane

of Fig. 1(a) [17]. The photons distributed in a conical beam shape with a field radius of 6 cm.

The energies were in the interval between 10 and 28 keV of Molybdenum $(Mo_{42})/$ Molybdenum (Mo_{42}) .



Figure 1: Geant4 geometrical model used to construct the electrons source in twodimensional (2D) with an angle of incidence of the anode at $\theta = 17^{\circ}$, as demonstrate in Figure 1 (a) and Figure 1 (b) that represent a schematic diagram of the beam geometry.



Figure 2: Virtual geometry layout of an X-ray tube with Molybdenum (Mo_{42}) target material through a Beryllium (Be_4) window arranged in Geant4.

Modelling the X-ray generation

The X-ray spectra collected where electrons emitted with energy of 50 keV at an angle of 90° towards the anode of molybdenum (Mo_{42}) . This accomplished by tracking a large number of incident electrons collide on the anode and then slowing down within the anode. This is because such electrons encounter attenuation in the anode material depending on the anode angle and the beam direction until they are absorbed or emerge from it. Then calculate the number of photons produced during their travel within the anode [18]. The anode is a tube made of Molybdenum (Mo_{42}) with a thickness of $4 \mu m$ and a 25 mm dimension. Its surface makes an angle of 17° with the beam line of the electron. A simplified geometry

configuration of the X-ray tube in a mammogram is shown in Fig.1 (b). Such geometry is the one that has chosen as input for the Geant4 code [19]. The X-rays photons propagate isotropically from the point of production, but with directions towards the Beryllium (Be_4) window. Such window was made of Beryllium (Be_4) box with a thickness of 0.5 mm but its dimensions was25mm. It was placed at 2 cm from the point where the electrons hit the anode [20] after passing through the (Be₄) window. Then the X-ray spectrum passes through filter material for absorbing the low energy photons. The filter was placed at a distance of 10 cm from the anode that made from 30 μ m Molybdenum (Mo_{42}) and air. This is to further attenuation of the X-ray beam before the measurement point [21]. Table1 summarizes all the X-ray tube parameters examined in this work.



Modelling lesion & Breast phantom:

Figure 3. Exemplar image of 5% and 50% glandular tissue phantom with 5mm tumor, the white indicates the region of the breast, and the region outside of the breast black, (a) and (b) contains a tumor in the center for 5% and 50% respectively. Fig. 3 (c) and (d) contains a tumor at the end of the breast phantom for 5% and 50% respectively

The modelled breast phantom was a semi-circular having a cylindrical shaped with a radius of 6 cm and a 4 cm thickness [22]. It consists of a homogeneous mixture of glandular and adipose breast tissue with a glandular fraction mass. It is worth pointing out that we have chosen a range of glandular fraction from 5% up to a 100%. Also we carried out investigations with no glandular fraction i.e. 0%. The materials of the chosen phantom mimic normal breast tissue with a density of 1.02 g/cm³ and adipose tissue with density of 2.0 g/cm³, respectively [23, 24]. The compositions of the different tissues taken from Geant4 [25]. The elemental compositions and densities of the glandular and adipose tissues defined and used according to Hammerstein *et al.* data [26]. After defining the normal breast tissues on the phantom, a tumor model added into it and placed in a particular position. Spherical particle with 5 mm in diameter made of breast tissue for a specific glandular part [27]. The

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tumour is located in two different positions within the breast phantom. The first position was in the center of the axis. Whereas the second position was at the plane below the breast with its central axis at the amorphous selenium (a-Se) detector [28], as shown in Fig.3 (a), and Fig.3 (b).

 Table .1: The X-ray mammography unit of the parameters introduced on the Geant4 to perform the simulation

Item	<u>Parameters</u>			
	Material	Thickness	Dimension	Position in System
Target	Molybdenum\(Mo ₄₂)	50 µm	25 mm	In the center
window	Beryllium (Be ₄)	0.5 mm	25 mm	2 cm away from target
Filter	Molybdenum (Mo ₄₂)	30 µm	25 mm	10 cm away from target
Detector	Amorphous selenium	250 μm	29x24 cm	60 cm away from target



Figure 4: Simulation system performed showing the Geant4 based mammography unit. *Contrast to Noise Ratio (CNR) measurement*

The CNR is a measure used to determine image quality. Defined by the ratio of the (average) pixel intensity difference between the detail signal and the background $(I_{Object} - I_{Bkgd})$, and the standard deviation of the background (σ_{Bkgd}) the CNR calculated was by using Eq.1: [29, 30]. This part is an attempt to provide a successful measurement of noise in simulated images using CNR.

$$CNR = \frac{l_{object} - l_{Bkgd}}{\sigma_{Bkgd}}$$
[1]

RESULTS AND DISCUSSION

Mammography unit Simulation

Detection of tumors in mammograms often limited by statistical variability of normal structure rather than image noise. This paper reports investigation of the statistical properties of tissue structures in X-ray projection mammograms, using simulated images. The goal is to understand statistical properties of breast tissue structure, and their effects on lesion detection. In the beginning, the spectra of the X-ray Mammography tube generated by mimicking a realistic clinical setting. To do this the Geant4 Code splatted into two parts to both reduce the length of the computational time as well as to increase the collected counts.

In the first part, the generation of the electron source inside the X-ray tube represented by electron beam with energy of 50 keV. Electrons shotted into Molybdenum (Mo42) anode target that placed at a distance of 10 mm. A window of Beryllium (Be4) with 0.5 mm thickness placed below the anode. After photons generated at the anode will then passed through a Beryllium (Be₄) window and then filtered by the (Mo_{42}) . Consequently, the detector records their energy. At first pure Molybdenum (Mo42) spectrum of the X-ray tube created. In the second part, the spectrum of photons extracted from the first code and then converted to probability density function. This help to generate photons as the primary particles in the second code using the inverse cumulative method. Then the phantom with the lesion simulated by recording on the imaging detector the energy and the positions information of photons. An amorphous selenium (a-Se) detector with a dimension of [29×24] cm², a thickness of 250µm and pixel size of 0.5 mm was the chosen X-ray mammography detector. The distance between the generation source of photon and the detection area was 60 cm. In the second part, the photon spectrum used as an input for the setup of the mammography unit simulation experiment. A phantom of 4 cm thickness with a variety of mammary gland densities were investigated. The main aim was to demonstrate the effects of the various mammary gland on tumour detection. In addition, its distribution in the projected images evaluated using CNR as shown in Figure 4.



Figure 5: Energy spectra of a 4μ m filtered a beam incident, transmitted through a thick phantom. The transmitted spectra includes contribution of a foil to the beam attenuation

Spectra analysis

Mammography has a low sensitivity in tumour detection particularly for women with dense breasts. This is due to low contrast between malignant and normal tissues confounded by the predominant water density of the breast. Water is found in both adipose and fibro glandular tissue and constitutes most of the mass of a breast. Tumors detection in mammograms limited by the marked statistical variability of normal structure rather than image noise. Characteristic X-rays produced when the electrons hit the anode material. If the energy of an incident electron exceeds the binding energy of the electron in the anode atom, so it is possible to eject the electron and ionize the atom [31]. This process creates a vacancy that filled with an electron from an outer shell. This process creates an X-ray photon with energy equal to the difference between the binding energy of the electron shells. Thus, the X-rays emit discrete energies that are characteristic of that element. These discrete energy peaks overlap on the continuous bremsstrahlung spectrum, as demonstrated in Fig.5 [32]. Bremsstrahlung radiation emitted when a high-speed electron passes close to the nucleus of the atom. Due to the positive charge of the nucleus, the high-speed electron slowed down releasing a Bremsstrahlung X-ray photon with part of its kinetic energy. In addition, the highspeed electron collides directly with the nucleus of the atom, where all its kinetic energy transformed into an X-ray photon [33]. Fig.5 shows the energy spectrum of Molybdenum (Mo42) anode tube with 2 characteristic peaks at 17.4 and 19.6 keV superimposed on the continuous bremsstrahlung spectrum at 28 keV with Mo_{42} anode and Mo₄₂ filtered tube arrangement. Two relatively small peaks observed near 6 keV. These peaks are due to K escape processes of the amorphous selenium (a-Se) detector, which related to the photoelectric absorption of the molybdenum fluorescent lines in the detector [34].

Energies required for mammographic imaging namely Bremsstrahlung and characteristic X-rays investigated. Photons not useful for imaging i.e. Bremsstrahlung X-ray photons below 17 keV and above 24 keV were excluded. Hence, minimizing it as it reduces the image contrast [35]. This achieved by increasing the *Molybdenum* (Mo_{42}) filter thickness. A spectrum produced by using Mo₄₂ anode tube at 28 keV. This filtered by 50µm of molybdenum as shown in Fig. 6. The filter removed almost completely from the X-ray beam after the K_{β} line of the Mo source (19.4 kV). It reduces the high-energy components at the same level of filtration. This resulted in an increase in the ratio of the photon 17-18 keV with respect to the total centred at 17.44 keV [36].

Figure 5 and Figure 6 showing the overall number of photons at different filter thicknesses within 10 - 20 keV energy range centred at 17.44 keV. Those were greater than the one obtained using a filter with a thickness of 50µm. Increasing the thickness of the filter will increase the number of photons from 600 to 700 counts.



Figure 6: Energy spectra of 50 μ m Mo42 with a 30 μ m Molybdenum filtered beam incident, transmitted through a phantom with 4 cm thickness. The transmitted spectra include contribution of a foil to the beam attenuation.

Table.2: Parameters for the molybdenum (Mo_{42}) Characteristic X-ray simulated usingGeant4.

Target	k_{α}/keV	k _β /keV
Molybdenum	17.48	19.61

Effect of glandular fraction distribution

A recent study [37] in a group of women showed a strong correlation between the glandular tissues appears on X-ray mammography and the detection of cancer. Such work based on classifying the regions of the breast into regions of lowest breast cancer risk when the glandular tissue is few and more dangerous when the breast contains a high glandular fraction. They concluded that increasing prevalence of glandular tissue in the breast rises the risk of cancer [38]. We also studied the breast density by varying glandular fraction according to the classification descried elsewhere [39]. The mammograms classified according to BIRADS criteria into four density categories. The upper bounds 25%, 50% 75% and 100% respectively. That is when referring to the breast at the highest level of glands. An extremely dense breast could obscure a lesion in a rates exceeded 74%. Highlights the lowest level of 24% [40], as demonstrated in Fig.7.







Figure.7: The categories describe the fraction of glandular tissue in the breast for mammography density classification and are (5%) top left, (25%) top center, (50%) top right, (75%) bottom left, and (100%) bottom center.

Image quality analysis

There are a number of difference in the composition of the breast tissue. The distribution of glandular tissue is asymmetric from the breast upper surface to the boundary between the central and surround the region of the breast. Through this feature, one can inferred the relative prevalence of these problems in the breast. By selecting two different Regions of Interest (ROIs), each given the symbols A and B respectively. A pixel size of $[2\times 2]$ at an equal distance from the images selected for all simulations of different breast density ranging from 5% up to 100%. This allows studying the variation of the mean value inside the ROI as a function of the distribution of the glandular fraction within the phantom along its propagation path. Lower values observed toward the base of the breast compared to its center. To verify this, five glandular distributions generated in the simulation. Starting from a low concentration of 5% to a highly concentration of 100% distribution. A CNR calculated between the breast that contains glandular fraction and the background for both regions A, B, as demonstrated in Fig. 6 & Table.1. The glandular distribution significantly affects the CNR parameters and values. When the CNR value increases significantly and this applies to the more concentrated glandular distribution. A glandular tissue has a higher CNR value than the less concentrated distribution, as depicted in Fig. 8 and Fig.9



Figure 8: An example of the distribution of a 2D breast phantom image with a diameter of 6 cm. Two ROIs are indicated (A, B), for assessing the quantitative prevalence of the glandular fraction in the breast phantom. Each ROI covers a region of pixels and A is position on the central axis of the phantom and B position at the edge of the axis



Figure 9: Contrast-to-noise ratio (CNR) measurement results in a 4 cm-thick breast phantom with different glandular fraction ratios, CNR is higher in glands <50 and CNR decreases with increasing mammary gland density in both regions.

Table 3:	The CNR	values	according to	a	glandular	fraction	in	the	breast	phantom	of
region A	and region	n B outl	ined in Figur	e 8	3						

Glandular fraction%	Contrast-to-Noise Ratio (CNR)			
	A B			
5	0.102	0.351		
25	0.05	0.292		
50	0.042	0.281		
75	0.033	0.272		
100	0.028	0.245		

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We also investigated the detection of the tumour inside the breast phantom. A tumour located within a region where glandular tissue localized and explored the phantom as well as in a less localized region. *ROIs* were drawn on each image, indicating the tumour region used. This is to determine the average signal intensities in the background. This refers to the breast region only. A square size of approximately [2x2] pixels chosen. For such work, a few images taken at glandular fraction of 5%, 25% and 50% in both modes. To detect the tumour a profile drawn on the images. Fig.10 and Fig. 11 showing profiles of the distribution along the white horizontal lines for a 4 cm thick breast phantom. A large variation noticed, but reversed in the sense that the values were lowest in the denser regions indicating the presence of tumour. For comparison, a profile drawn for each image of the phantom with a tumour at the center or at the end of the phantom with glandular fraction of 5%, 25% and 50%. An increase in the tumour signal noticed.



Figure 10: A comparison is provided for all the horizontal profiles along a diameter 6 cm of the breast phantom with a 5 mm diameter tumour in the center at (a=5%)

In addition, the shape clearly defined with a tumour at the end of the phantom. Moreover, the increase in background noise is also evident as shown in Fig. 11 and Fig. 12 that demonstrate the difference between tumour tissue signal and the surrounding structures. It increases in peripheral regions in which glandular tissue prevalence decreases, leading to an increase in CNR and ease of tumor detection. A high contrast for the area under the breast (least dense) 6.69 compared to the central region greatest density i.e. 50% glandular fraction. Accordingly, the smaller the glandular part, the more the lump became apparent. The tumor becomes evident in areas less than 50% and is difficult to detect after that as the areas thicken and obscure the tumor.



Figure 11: A comparison for the horizontal profiles along a diameter 6 cm of the breast phantom with a 5 mm diameter tumour in the center at (b=25%, c=50%) and at the end at (d=5% of the breast. Profiles taken in a same manner demonstrated in Fig. 10(a).



Figure 12: A comparison for the horizontal profiles along a diameter 6 cm of the breast phantom with a 5 mm diameter tumour at the lower part (end or edge) of the phantom at (f=25%, g=50%) of the breast.

The effect of glandular fraction on linear attenuation coefficients:

The measurement of the linear attenuation coefficients of breast tissues is of fundamental importance in breast X-ray diagnostic imaging. The tumour protrudes when placed in less invasive regions of mammary glandular tissues. It was difficult to detect small tumours placed in larger content of mammary gland. A CNR comparison suggested the CNR rate decreased when imaging a small tumour size placed in a phantom with larger mammary gland content. Otherwise, its height at low values the mammary gland, the detection ability of the tumour decreased. This attributed to the linear attenuation coefficients of the tumour and the mammary gland that are very close to each other. If the mammary gland formation increases, it is difficult to detect tumour and the contrast between them decreases.

Table.4: Linear attenuation coefficient properties of breast tissue and glandular at Mo_{42}/Mo_{42} . The energy is in the interval between 10 and 28 keV.

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glandular fraction%	Attenuation Coef $\mu_{\text{glandular}}[cm^{-1}]$	$\mu_{Breasttissue}$		
	high prevalence region	low prevalence region	[<i>cm</i> ⁻¹]	
5	0.94135	0.94082	0.1163	
25	0.94084	0.94130	0.1164	
50	0.94076	0.94129	0.1167	
75	0.94099	0.94103	0.1172	
100	0.94053	0.94123	0.1177	

Result in Fig. 13, demonstrate and explain the relationship between the linear attenuation coefficient/cm and the glandular fraction of the breast phantom. Noting that the tumor placed on two sites at energy in the interval between 10 to 28 keV. The breast density curve describes in four groups: 1) less than 25%, 2) 25% to 50%, 3) 50% to 75%, and 4) greater than 75%. The largest differences in attenuation observed at glandular fraction less than 25%. We were able to differentiate between them very clearly. However, greater than 50% glandular fraction were difficult to distinguish. A good agreement of glandular tissue found at a ratio between 25% and 50% in both tumor sites as demonstrated in Table.4. Such results established that mammographic density reduces the detection and mask out the tumour. In a condition greater than 50%, the tumor was missing on the image but detected in a condition that was less than 25% glandular fractions. Results confirmed the observed strong non-linear relationship between the classification of breast density and tumor detection. These where obvious and demonstrated clearly in Fig. 11 and Fig. 12. Breast density has a definite impact on sensitivity of breast cancer detection. This may help to predict a woman's future risk of breast cancer.

DISCUSSION

In this study, the X-ray mammography tube modelled and performed in a similar manner described by Nasseri, M. M. [10]. The spectra of the Molybdenum anode (Mo_{42}) target, that used with a 30 $\mu m Mo_{42}$ filter in a Mo_{42}/Mo_{42} combination gives the characteristic X-ray peaks of Mo. In other words, $k_{\alpha} = 17.48 keV$ and $k_{\beta} = 19.61 keV$ clearly identified and shown in Fig. 3 and Fig.4. A comparison was carried out by using two different filters with a thickness of 30 μm and 50 μm but at a fixed energy of 28 keV. The X-ray photon intensity displayed a higher value when using a thickness of 50 μm with characteristic peaks displayed but most of the low-energy rays were removed from the image. The simulation also included the breast phantom either with tumour inside it or without tumour. Images obtained one by one while varying glandular fractions and implementing the acquisition protocols to mammogram. The simulated mammography imaging systems calculate the distribution of glandular fraction as a

function of the breast density. The two-dimensional (2D) image, the contrast value, the curve relating glandular fraction to the CNR were investigated.



Figure 13: The relationship between the X-ray linear attenuation coefficient of the breast tissue and glandular tissue versus glandular fraction%. (a) Contains a tumor in the center, (b) contains a tumor at the end of the breast.

Then the effect of the image contrast of breast density calculated for all different ratios of glandular fraction. The calculations obtained in two different regions. The first region is at the center of the breast phantom and the second region at the edge region of the phantom. Table 3 outlined the calculated CNR values for five different ratios discussed in the Fig.8. For each ratios glandular fraction i.e. 5%, 25%, 50%, 75% and 100% the CNR versus plotted. Breast density of glandular fraction values at 5%, 25%, 50%, 75% and 100% on the two regions demonstrated in the Fig.9. Interestingly, our results show variations of CNR values within the breast tissue model for the two regions. They were low at high glandular fraction of 50% given values A=0.042 and B=0.281. It increases with the decrease in glandular fraction of 5% given different values A=0.102 and B=0.351. This suggests that breast density might be a strong predictor of CNR, independent regardless of the position tumors.

On the other hand, this work confirmed that the main contribution to tumour detection comes from the effect of the profile obtained from the two tumour regions as shown in Fig.10. Also showed a clear variation in a glandular fraction of less than 50%. The difficulty in detecting a

tumour in the phantom with a density greater than 50% is because the linear attenuation coefficients of the tumour and the glandular tissue are close to each other. As shown in Fig.10, Fig. 11, Fig.12 and table.4, the increase in density obscures the tumour. This result in the difficulty of detecting the tumour, these results corroborate previous study [42] that demonstrated reduction in cancer diagnosis sensitivity and poor image quality in high-density mammary glandular fraction. Our results indicate that the CNR used as an indicator for assessing diagnostic reliability in breast cancer to control the effect of the mammary gland ratio on image quality. This may improve the reliability of diagnosis, which is on dependent on the mammography readers. Our findings confirmed previous studies [43, 44] that show and concluded the effect of breast density on breast cancer diagnosis.

CONCLUSIONS

The entire cone-beam for detecting a small size tumor at low energies mammography modelled and described. Results were in good agreement with those reported by others authors, who had used Monte Carlo simulation to model the X-ray mammography tube [18-21]. It also shows an inverse relationship between the detection of tumors at different densities (glandular fractions) on mammogram. The effect of this on the quality of the produced images demonstrated using CNR. The simulation performed with a fixed X-ray tube of a cone-beam projecting perpendicular based on a parametric representation of the breast size at a distance of 60 cm from the detector. The photons distributed at an angle of 12° with a field radius of 6 cm at energy in the interval between 10 and 28 keV of Mo / Mo.

An anthropomorphic breast phantom with a thickness of 4 cm containing a 5 mm tumour modelled. The glandular tissue distributed at an asymmetrical concentration of 5%, 25%, 50%, 75% and 100% were considered.

Five images of breast phantom acquired at different breast densities without tumour in placed. The CNR for a region in the center and at the end part (edge) of the breast were calculated and reported. Another five images of the same breast phantom but with a 5 mm tumour in placed also obtained. Very low-contrast embedded in dense glandular tissue center, and in a phantom edge. Then a calculation of the CNR and comparative studies between all cases carried out and reported. Our results indicated that the breast with less density gets higher contrast and that under the same beam conditions; the breast with larger density gets less contrast. The distribution of glandular fractures within the breast volume during irradiation plays an important role in the diagnosis, where the task is to detect this tumour. The difficulty of distinguishing between the tumor tissue and the glandular fraction in the breast was due to the problem of breast tissue overlapping. This is the one of the main factor in the formation of breast density.

Clinically the malignant processes resulting from X-rays originate in the glandular tissue only. The quantity and the spatial distribution of tissues is quite variable among women. Women with dense breasts combined with an established risk of developing breast cancer may benefit from additional imaging methods because X-ray mammography alone is not sufficient for early detection of breast cancer. Individual breast cancer risk assessment is critical to allow clinicians to recommend appropriate screening strategies for their patients and to highlight breast density and overall risk of developing breast cancer.

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