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Visibility of microcalcification in cone beam breast CT – Effects of x-ray tube voltage and radiation dose

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Abstract

Mammography is the only technique currently used for detecting microcalcification (MC) clusters, an early indicator of breast cancer. However, mammographic images superimpose a three-dimensional compressed breast image onto two-dimensional projection views, resulting in overlapped anatomical breast structures that may obscure the detection and visualization of MCs. One possible solution to this problem is the use of cone beam computed tomography (CBCT) with a flat-panel (FP) digital detector. Although feasibility studies of CBCT techniques for breast imaging have yielded promising results, they have not shown how radiation dose and x-ray tube voltage affect the accuracy with which MCs are detected by CBCT experimentally. We therefore conducted a phantom study using FP-based CBCT system with various mean glandular doses and kVp values. An experimental CBCT scanner was constructed with a data-acquisition rate of 7.5 frames/s. 10.5- and 14.5-cm-diameter breast phantoms made of gelatin were used to simulate uncompressed breasts consisting of 100% glandular tissue. Eight different MC sizes of calcium carbonate grains, ranging from 180–200 μm to 355–425 μm , were used to simulate MCs. MCs of the same size were arranged to form a 5 $\times$ 5 MC cluster and embedded in the breast phantoms. These MC clusters were positioned at 2.8 cm away from the center of the breast phantoms. The phantoms were imaged at 60, 80, and 100 kVp. With a single scan (360 degrees), 300 projection images were acquired with 0.5 $\times$, 1 $\times$, and 2 $\times$ mean glandular dose limit for 10.5-cm phantom and with 1 $\times$, 2 $\times$, and 4 $\times$ for 14.5-cm phantom. Feldkamp algorithm with a pure ramp filter was used for image reconstruction. The normalized noise level was calculated for each x-ray tube voltage and dose level. The image quality of CBCT images was evaluated by counting the number of visible MCs for each MC cluster for various conditions. The average percentage of the visible MCs were computed and plotted as a function of the MGD, the kVp, and the average MC size. The results show that the MC visibility increased with the MGD significantly but decreased with the breast size. The results also show the x-ray tube voltage affects the detection of MCs under different circumstances. With a 50% threshold, the minimum detectable MC sizes for the 10.5-cm phantom were 348 ($\pm$ 2), 288 ($\pm$ 7), 257 ($\pm$ 2) μm at 3, 6, and 12 mGy respectively. Those for the 14.5-cm phantom were 355 ($\pm$ 1), 307 ($\pm$ 7), 275 ($\pm$ 5) μm at 6, 12, and 24 mGy, respectively. With a 75% threshold, the minimum detectable MC sizes for the 10.5-cm phantom were 367 ($\pm$ 1), 316 ($\pm$ 7), 265 ($\pm$ 3) μm at 3, 6, and 12 mGy, respectively. Those for the 14.5-cm phantom were 377 ($\pm$ 3), 334 ($\pm$ 5), 300 ($\pm$ 2) μm at 6, 12, and 24 mGy, respectively.

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Keywords

cone-beam computed tomography; breast imaging; flat-panel detector; microcalcifications; mean glandular dose

I. INTRODUCTION

Breast cancer represents a major public health problem. According to the American Cancer Society, breast cancer is the second leading cause of cancer death in women today, and more than 210,000 U.S. women will develop breast cancer in 2006.¹ Women have a gradually increasing risk of developing breast cancer during their lifetime, especially for those age 50 years and older. Thus, early detection of breast cancer is very important and is one of the major reasons that the breast cancer mortality rate has declined at an average of 2.3% per year between 1990 and 2002.

For the detection of early breast cancer, x-ray mammography is the most common imaging technique, and mammography is a commonly used tool in diagnostic evaluations. Mammography is currently the best technique for detecting microcalcifications (MCs), which may represent early breast cancer. Therefore, improving detection and visualization of MCs on mammograms is very important in breast cancer screening and diagnosis. For the past two decades, many studies have investigated anti-scatter techniques, x-ray spectra, and radiation dose in order to improve image quality for screen/film and digital mammography.^{2–8} However, despite these efforts, an ongoing problem with mammographic images is that they superimpose a three-dimensional (3D) compressed breast image onto two-dimensional (2D) projection views, resulting in overlapping anatomical structures that may, depending upon their size and location, cause MCs to be obscured on mammography, even when the MCs have an adequate contrast-to-noise ratio (CNR). On mammography, these anatomical structures are seen because of the differences in the x-ray attenuation between adipose tissue, glandular tissue, ducts, vessels, and soft tissue masses in the breast. Approximately 15% of breast cancers are not identified with mammography. Also, approximately 65% of lesions suggestive of breast cancer are benign on biopsy.^{9, 10}

One of possible solutions to the problem of overlapping anatomical structures on mammography is cone beam computed tomography (CBCT) acquired with a flat panel (FP) digital detector. Computed tomographic techniques were introduced and applied to breast cancer in the mid-1970s,^{11, 12} but little progress was made because of limitations such as low spatial resolution, long scanning time, large slice thickness, and high radiation dose to the patients. In recent years, various digital detectors have been developed and commercialized, including computed radiography systems,^{13–15} charge-coupled device-based systems,^{16–18} and amorphous silicon FP-based systems.^{18–21} These digital detectors use different techniques to improve detective quantum efficiency and provide several advantages, including wide dynamic range, excellent linearity, reasonable spatial resolution, and the absence of geometrical distortion.

With the advent of the FP digital detectors, a number of studies have been conducted to investigate the feasibility of CBCT for breast imaging. Boone *et al.*²² developed the concept of a dedicated CT scanner for breast imaging and reported that the CT dose for a single scan was comparable to that for a two-view mammographic examination (per breast) for 5-cm thick compressed breasts and that the dose was less than that needed for two-view mammography for thicker breasts. Ning *et al.*²³ imaged phantoms with a CBCT system to evaluate image quality and reported that, with a total dose of less than 4 mGy, a CBCT system can detect a 1-mm carcinoma and a 200- μ m calcification in 11-cm diameter uncompressed breasts. Gong *et*

*al.*²⁴ conducted a computer simulation to investigate the ability of a FP-based CBCT system to detect MCs and reported that 175- μ m MCs could be detected with a total dose of 4 mGy. In addition, Gong *et al.* compared digital mammography and a FP-based CBCT system for detecting 5-mm computer-simulated masses and reported that a CBCT system performed significantly better than digital mammography.²⁵ These studies demonstrated that CBCT breast imaging reduces the anatomical noise from overlapping structures effectively and significantly improves the detectability of small lesions.

These results were promising. However, these studies did not show experimentally or quantitatively how radiation dose affects the accuracy of CBCT in detecting MCs. To address this issue, we conducted a phantom study using a FP-based CBCT system at various radiation doses. Although there is also a need to evaluate detector pixel size, x-ray spectrum, and breast composition, radiation dose and image quality are fundamental aspects of a dedicated CBCT system, especially in the detection of MCs for screening. Thus, we believe that this investigation represents a necessary first step in assessing CBCT breast imaging.

II. MATERIALS AND METHODS

A. Experimental FP-based CBCT system

An experimental FP-based CBCT system was built on the optical bench in our laboratory (Fig. 1, left). The system consisted of a general radiographic x-ray unit, an amorphous silicon/cesium iodide (a-Si:H/CsI) FP detector, and a step motor-driven rotating table. The x-ray unit is composed of an x-ray generator (Indico 100 SP, Communications & Power Industries, Ontario, Canada) coupled to an x-ray tube (G-1592, Varian Medical Systems, Salt Lake City, UT) with a nominal focal spot size of 0.6 mm. The operating range of the x-ray unit is up to 150kV and 125 kV for radiographic and fluoroscopic use, respectively. The FP detector (Paxscan 4030CB, Varian Medical Systems) consists of a CsI(Tl) scintillator coupled to an amorphous silicon photodiode array. The matrix of the detector is 2048 $\times$ 1536 pixels with 194 μ m pixel pitch, resulting in 40 $\times$ 30 cm² active area. In addition, the FP detector can be configured as a binning mode (2 $\times$ 2) with 388 μ m-pixel. The detector has capability of data acquisition rates of 7.5 and 30 frames/sec (fps) for the non-binning and binning modes, respectively. A holder was used to position phantoms and to simulate a pendant breast. Several different size holders were available for various breast sizes. The holder was then placed on the rotary table driven by a step motor (B4872TS, Velmex Inc, Bloomfield, NY) to rotate the breast phantoms.

The focal spot plane was aligned at the middle plane of the FP detector. The x-ray source-to-isocenter distance (SCD) was 75 cm, and the distance between the x-ray source and the FP detector (SID) was 100 cm. Therefore, the magnification, defined as the ratio of the SID to the SCD, is 1.33 in this study.

B. Selection of kVp settings

In CBCT breast imaging, the breast undergoes no compression, which results in more attenuation of x-rays as they travel through thicker breast tissue than is encountered in conventional x-ray mammography. Therefore, it might be expected that higher kVp values may be more appropriate to be used for CBCT breast imaging compared to mammography. Several studies^{24, 26, 27} using different approaches to determine the optimal kVp setting for CBCT breast imaging have reported that the optimal operating range is between 50 and 70 kVp for a breast size of approximately 11 cm in diameter to obtain a maximize CNR. In addition, Boone recommended use 80 kVp to image various breast sizes.²⁸ Therefore, 60 and 80 kVp were selected. Furthermore, higher kVp may be needed for larger breasts, so that 100 kVp was selected for this study to investigate MC visibility as well. In this study, no additional filter was used. The beam quality of x-ray beam was then measured by performing half-value layer

(HVL) measurement for 60, 80 and 100 kVp using 1100 aluminum. An ion chamber with 150 cc (model 96035B, Keithley Instruments, Cleveland, OH) and a dosimeter (model 35050A, Keithley Instruments) were used. The HVLs were found to be 3.17 mm, 4.27 mm, and 5.62 mm Al for 60, 80, and 100 kV, respectively.

C. Simulated breast phantoms and CT number

In order to simulate pendant geometry for an uncompressed breast proposed by Boone et al.²², two cylindrical breast phantoms made of gelatin were used in this study. The dimensions of these breast phantoms were 10.5 cm (Fig. 1, right, referred to as small breast phantom in the rest of the paper) and 14.5 cm (referred to as large breast phantom in the rest of the paper) in diameter, respectively, and 12 cm in height for both breast phantoms.

Prior to determining the property of the simulated breast phantoms used in this study, the linear attenuation coefficient (LAC) of water was measured by using a customized water-equivalent phantom with a dimension of 15 cm in diameter (Catphan, The Phantom Laboratory, Salem, NY). The water phantom was scanned with a collimated x-ray beam with slot width of about 100 pixels at 60, 80, and 100 kVp for a 360-degree rotation at 7.5 fps data acquisition rate. Because beam hardening occurs increasingly close to the center of the phantom, eight locations close to the periphery of the water phantom were selected to calculate the water's LAC, each consisting of 1200 voxels. The mean value of the water's LAC was then averaged over eight locations and was found to be $0.2535 \pm 0.0011 \text{ cm}^{-1}$, $0.2328 \pm 0.0012 \text{ cm}^{-1}$, and $0.2228 \pm 0.0009 \text{ cm}^{-1}$ for 60, 80, and 100 kVp, respectively. Repeatedly, the large breast phantom (similar dimension to the water-equivalent phantom) was scanned at 60, 80 and 100kVp, and the water's LACs were applied to calculate the CT numbers for the simulated breast phantoms. It was found that the CT numbers of the gelatin were 217.4 ± 5.1 , 185.1 ± 7.7 , and 169.7 ± 6.6 for 60, 80, and 100 kVp, respectively. Meanwhile, the CT numbers for the fat and fibrous tissues were calculated based on the linear attenuation coefficients measured by Johns and Yaffe.²⁹ It was found that the CT numbers were -240 and 35 for fat and fibroglandular tissues, respectively, for x-ray spectrum with 80 kVp. These results indicate that the simulated breast phantoms are more likely glandular tissue.

D. Simulated microcalcification phantom

Eight different sizes of calcium carbonate grains (Computerized Imaging Reference Systems, Norfolk, VA) ranging from 180–200 μm to 355–425 μm were used to simulate MCs. These MCs are irregular shapes and were siphoned into different size groups. The MCs with the same size group were arranged to form a 5×5 MC cluster. The spacing between two adjacent MCs within each individual MC cluster was 3 mm. These eight MC clusters were positioned at the same slice planes for both breast phantoms. The centers of individual MC clusters were 2.8 cm away from the center of the breast phantom for both breast phantoms. Figure 2 shows the layout (left) of the MC clusters and one reconstructed CT slice of MC images (right) for large breast phantom.

E. Exposure measurement and dose calculation

The mean glandular dose (MGD) could be estimated by the following equation:

$$D = D_{gN}(CT) \times E \quad (1)$$

where D is glandular dose expressed in milligrays (mGy), E is the measured total exposure at the isocenter of the CBCT system expressed in roentgens (R), and $D_{gN}(CT)$ is the normalized dose conversion factor and is defined as the ratio of the glandular dose in the breast to the air kerma at the isocenter of the CBCT system with the unit in mGy/R. $D_{gN}(CT)$ is a function of

the x-ray spectrum and the breast size and tissue composition, and is independent of SCD. Based on the previous measurement of the CT number for the breast phantoms, the $D_{gN}(CT)$ values for 100% glandular tissue reported by Boone *et al.*²⁸ were used in this study.

The exposure rates were then measured at the isocenter of the CBCT system with a pencil probe ion chamber (10X5-6, Radcal Corp., Monrovia, CA) in air with the phantoms removed while the data acquisition rate was 7.5 fps for x-ray tube potentials at 60, 80, and 100 kVp. Because the scanning time for each single 360-degree rotation was 40 seconds, the exposures for each scan were computed by multiplying the exposure rate and the scanning time. By adjusting the tube current (mA) and the pulse duration of x-ray beam (10 msec and 30 msec), and, according to Eq. (1), by multiplying $D_{gN,100}(CT)$ values, imaging techniques were determined (Table 1) for air exposures to obtain the MGD equal to 3, 6, and 12 mGy for small breast phantom except 3mGy at 100 kVp, and 6, 12, and 24 mGy for large breast phantom except 24 mGy at 60 kVp. The air exposures were 1.01, 0.87, and 0.83 R/rotation at 6 mGy for 60, 80, and 100 kVp, respectively, for small breast phantom, and 2.4, 1.94, and 1.79 R/rotation at 12 mGy for large breast phantom. According to the Mammography Quality Standards Act,³⁰ the limit of the mean glandular dose (MGD) for a two-view x-ray mammography examination is 6 mGy for each breast. Using this limit as reference, the glandular doses of 3, 6, and 12 mGy for small breast phantom were referred to as 0.5×, 1×, and 2× of the MGD in this manuscript, respectively. Similarly, the glandular doses of 6, 12, and 24 mGy for large breast phantom were referred to 1×, 2×, and 4× of the MGD.

F. Image acquisition and reconstruction

The sequence of projection data was acquired by initializing pulsed beam to trigger readout synchronously while data acquisition rate was 7.5 fps with non-binning mode. During data acquisition, the breast phantoms were rotated over 360 degrees for each single scan. With each scan, 300 projection images were acquired. Then, Feldkamp's backprojection algorithm with a pure ramp filter was used for 3D reconstruction. The data acquisition and reconstruction parameters are listed in Table 2. The breast phantoms were imaged three times for each condition. Because all MC clusters were placed at the same CT slices and same distance away from the phantom centers for both breast phantoms, no attempt was made to correct for the cupping effect.

G. Observer study and statistical analysis

Thirty 800×800 slices were extracted from the reconstructed image data (50–1100×1100 slices) for each MGD level and x-ray tube voltage for individual breast phantoms. As a result, forty-eight CT image sets were generated. The MC clusters were then circled by solid rings on all slices to indicate the MC locations. The CT images were displayed sequentially using a software program (MRIcro) on a 1600×1200 CRT monitor and reviewed independently by five physicists. The reading order of the CT images was randomized for each observer. The observers were allowed to alter the window and the level settings and to step through the slices back and forth. Also, the observers were asked to score the numbers of visible MCs for individual MC clusters for each CT image set. No constraints on time or viewing distance were imposed.

The recorded numbers of visible MCs from the individual observers were categorized by the MC sizes, the dose levels and the x-ray tube voltages, and then computed to obtain the percentage of the visible MCs for each condition. The percentages were averaged (referred to as the percentage visibility in the rest of the paper) across over the five observers and were plotted as a function of the MGD for eight MC sizes for both phantoms. The percentage visibilities were also plotted as a function of the x-ray tube voltage for eight MC sizes for both phantoms. Furthermore, the percentage visibilities were plotted as a function of the average

MC size for both breast phantoms in this study and used as a quantitative measure for the MC detectability. Student *t*-test was used to compute the *p*-values to quantify the statistical significance for performance differences in the percentage visibilities between dose levels and between x-ray tube voltages.

H. Image noise measurement

The noise property was measured on the reconstructed CT images as basis of the CT numbers. The background (*b*) and the standard deviation (σ) of the breast phantoms were measured at the same locations of these eight MC cluster centers but different slices (z-direction) where no MCs were included for each dose level and x-ray tube voltage. Then, the mean background (b_{mean}) and mean standard deviation (σ_{mean}) were calculated across over these eight locations by using following equations

$$b_{\text{mean}} = \sum_{i=1}^N b_i / N \quad (2)$$

$$\sigma_{\text{mean}} = \left(\sum_{i=1}^N \sigma_i^2 / N \right)^{1/2} \quad (3)$$

where *N* is 8 in this study. The mean standard deviation was then divided by the mean background and plotted as a function of MGD for different x-ray tube voltages for both phantoms.

III. RESULTS AND DISCUSSION

Figure 3 shows six consecutive slices (7th – 12th) of reconstructed MC images acquired at 80 kVp with an MGD of 12 mGy for the small phantom. In these images, 250–280, 280–300, and 355–425 μm MCs were visible, but 300–355 μm MCs were not visible because they were captured in the 21st – 24th slices. The reason that MCs were not visible at the same slices because it was difficult to align all MCs together during phantom fabrication and to position the breast phantoms for cone beam scanning.

A. Effect of the MGD

Figures 4(a) and (b) show the percentage visibility plotted as a function of the MGD for various MC sizes at 60, 80, and 100 kVp for the small and large phantoms, respectively. As expected, for each kVp, the percentage visibility increases with the MGD for several MC size groups. This can be explained by the fact that the noise in the reconstructed CT image decreased with the MGD (normalized noise level used in Figure 5) for all kVp settings and both breast phantoms. In addition, the MC contrast on the CT images is the difference in the LAC between

MCs and the breast phantom relative to the LAC of water, $C = (\mu_{mc} - \mu_b) / \mu_w \times 1000$, where *mc*, *b*, and *w* referred to the MC, breast tissue, and water. As a result, the contrast-to-noise ratio (CNR) of MCs in the CT images is proportional to the square root of the MGD,³¹ indicating that the higher the MGD, the higher the CNR, and thus, the higher the percentage visibility.

There are several key observations in Figures 4. First, the percentage visibility for 224–250 μm MCs or smaller were lower than 20% and varied little with the MGD and the kVp. This may be attributed to the spatial resolution limit of the system imposed by focal spot blurring

and image signal spreading in the detector. The cone beam CT system was built with the magnification factor of 1.33, which indicates an object at the isocenter would be blurred to a size of about 200 μm due to focal spot blurring, similar to the detector pixel size. Additionally, the MTF of the detector were reported $>45\%$ and $\sim 20\%$ at the spatial frequency of 1.0 and 2.6 mm^{-1} (Nyquist frequency), respectively, for non-binning mode.^{32, 33} This indicates that the detector may not distinguish objects smaller than 385 μm on the detector plane. This result may reflect the limit of object size in our system being about 280 μm . Therefore, these blurring effects may prevent the MCs from being visible no matter how much exposure was used. Second, when larger MCs were imaged (such as 250–355 μm), the MC visibility increased with the MGD. This variation may be attributed to the limit of the quantum noise. Since the noise level in the reconstructed images decreases with the MGD, the MC visibility generally improves with the MGD, as long as MCs are not too small to be limited by spatial resolution of the system. According to the above concerns, the MCs ranging from 250–280 to 300–355 μm were selected to compute the percentage visibilities for various combinations of the kVp setting and the MGD (Table 2). The percentage visibilities were then used to compare the detection performance between different MGD (such as 3 mGy vs. 6 mGy or 6 mGy vs. 12 mGy) for given kVp settings. It was found that the percentage visibilities at high MGD's were statistically significant better ($p < 0.01$) than those at the low MGD's for all kVp's and breast sizes.

B. Effect of the x-ray tube voltage

Figures 6(a) and (b) show the percentage visibility plotted as a function of the x-ray tube voltage for various MC size groups and MGD's for the small and large phantoms. In Table 3, the average percentage visibilities for different kVp's with the same MGD are shown in the same row for comparison, and the p -values were computed and listed in Table 4. For the small phantom, it was found that the percentage visibilities in 60 kVp images were significantly higher ($p < 0.01$) than those in 100 kVp images at 6 and 12 mGy. The percentage visibility in 80 kVp images was significantly higher ($p < 0.01$) than that in 100 kVp images at 6 mGy. No statistically significant difference was found for percentage visibilities between 60 kVp and 80 kVp for all MGD values. For the large phantom, no statistical significance was found for percentage visibilities between different kVp settings for all MGD values except that 60 kVp and 80 kVp performed significantly better ($p < 0.01$) than 100 kVp at 12 mGy.

The results showed that 60 kVp and 80 kVp had similar performance and were better than 100 kVp for the small breast phantom while MGD was 6 mGy and for the large breast phantom at 12 mGy. This can be explained by the following observation. The CT numbers of simulated MCs were found about ranging from 1000–1300 (for all kVp's). Those of the simulated breast phantoms (made of gelatin) were about 170–220 (Section II.B.), resulting in MC contrast ≥ 800 , which is high enough. Thus, the detection of MCs in CBCT images may be affected by x-ray tube voltage but not limited by the contrast. Instead, the detection of MCs in CBCT images may be affected by the MGD and blurring.

There is an important observation for this result. It is well known that x-ray tube and detector motion can cause blurring so that using pulsed x-ray beam, instead of continuous x-ray beam, can reduce the effect. Table 1 shows that 30-msec pulsed beam was used for all CBCT images acquired with 60 kVp but fewer images with 100 kVp. The blurring effect can be estimated quantitatively for our study. Because the radius between the isocenter and center of each individual MC cluster is 2.8 cm, the blurring is about 44 and 133 μm for 10- and 30-msec pulsed beam, respectively. This indicates that the blurring for 30-msec pulsed beam is similar to the pixel size after demagnification but can be ignored for 10-msec pulsed beam. The results showed that CBCT images acquired with 60 kVp performed better than those with 100 kVp. This implies that the blurring due to x-ray tube and detector motion may not be a crucial factor

to affect the detection when the objects are located at certain distance from the isocenter (for example, 3.1 cm in our study). Further study is necessary to investigate the blurring effect along radial distance.

C. Percentage visibility vs. MC size

Figures 7(a) and (b) are two examples showing the percentage visibility in 80 kVp images plotted as a function of average MC size for three MGD values for the small and large phantoms, respectively. It is well known that measured probability of correct response in receiver operating characteristic or alternative force-choice studies can be fitted to a Gaussian integral model.^{34, 35, 36} A similar model was employed to fit the percentage visibility versus MC size data for different combinations of MGD, kVp and breast size using the maximum likelihood estimation. The function used may be expressed as:

$$P(x) = P_N + (1 - P_N) \times \int_{-\infty}^x \frac{e^{-\frac{(s-\mu)^2}{2\sigma^2}}}{\sigma \sqrt{2\pi}} ds \quad (4)$$

where the parameters x , P_N , μ , and σ^2 represent the input variable of MC size, the random probability due to the spatial resolution limit and image noise, the MC size at which the slope of the curve is maximum, and the noise variance on behalf of the steepness of the function, respectively. It was observed that the fitting curves shifted toward the left (smaller MC size) with increased MGD at all kVp's, indicating the possibility that smaller MCs may be detected if higher exposures are applied. For a given threshold visibility, the fitting curves may be used to determine the minimum detectable MC size. A threshold value of 50% or 75% corresponds to the steepest region of the curve and therefore may be used to more consistently determine the minimum detectable MC size. Table 5 lists the minimum detectable MC size for various combinations of the MGD, kVp, and breast size determined with the two different threshold levels. It shows that there is no statistically significant variations with the x-ray tube voltages for a given MGD but significant variations with the MGD for a given x-ray tube voltage for both phantoms.

The minimum detectable MC sizes were found to increase significantly with the breast size at the same MGD (288 to 355 at 6 mGy and 257 to 307 at 12 mGy using 50% as the threshold). This may be attributed to the difference in x-ray attenuation and scatter between small and large phantom. Both primary and scattered x-rays contribute to absorbed dose. However, scattered x-rays increase the noise level without contributing to the attenuation information in the reconstructed images. When the breast diameter increased from 10.5 cm to 14.5 cm, exposure was varied to keep the MGD the same. As a result, a larger fraction of the MGD originated from the scatter radiation and a smaller fraction from the primary radiation. The noise level was therefore increased, leading to lower MC visibility.

D. Breast density

Another important issue that may affect MC visibility is the breast density (composition). In this study, gelatin was used to simulate dense breasts. In sections II.B and III.B, we reported that the CT numbers of fat tissue, glandular tissue, and MCs are about -240, 35, and ≥ 1000 , respectively. As a result, the MC contrasts in adipose and fibroglandular tissues are ≥ 1240 and > 950 , respectively, indicating that MCs are a high contrast object in either fatty or dense breasts. Besides, the MC contrast in the breast phantom (gelatin) is ≥ 800 (Section III.B.). It means that this study provides more challenging imaging situations (due to lower contrast because denser breast phantoms were used), which is also consistent with the fact that MCs are usually present in fibroglandular tissue rather than adipose tissue. Given the higher contrast between MCs and

adipose tissue or MCs and glandular tissue, the minimum detectable MC sizes are expected to be lower than our measurement results for MC detection in adipose and glandular tissues.

Geometric alignment is another source for degradation of the spatial resolution. Accurate geometric alignment is essential to accurate image reconstruction to preserve correct object locations, shapes and sizes and to optimization of the spatial resolution. To check the system alignment, we usually image calcium carbonate grains placed at selected locations on a stereofoam block or aluminum wires suspended in air. The high contrast between calcium and essentially air provides artifacts that may indicate the type and degree of misalignment of the system.

There are several limits for this study. First, the MGD was calculated based on the $D_{gN}(CT)$ values from Boone's data, and $D_{gN}(CT)$ depends on x-ray spectra (HVL), breast size and density. In this study, the HVL of x-ray beam was reported as 4.27 mm Al at 80 kVp compared to 5.3 mm Al for Boone's model,²⁸ implying that the x-ray spectra used in this study may not be similar to what Boone used for the dose estimation. Second, the density of the breast phantoms used in this study is denser than 100% glandular tissue, such that no $D_{gN}(CT)$ values were available for these phantoms. Under this circumstance, the use of the $D_{gN}(CT)$ values from Boone's data may not be able to obtain accurate MGD values.

CONCLUSION

We conducted an experimental imaging study to investigate the effects of radiation dose and x-ray tube voltage in the detection of MCs in CBCT images. We performed a perception study on the visualization of MCs in the reconstructed images. The results indicated that the MC visibility increased with the MGD significantly but decreased with the breast size. The results also show the x-ray tube voltage affects the detection of MCs under different circumstances. With either 50% or 75% threshold, the minimum detectable MC sizes were not statistically significant between the x-ray tube voltages but significant variations with the MGD for both breast phantoms.

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Figure 1. Experimental stationary gantry CBCT system (a) with a breast phantom (b) placed on a rotating table.

(a)

(b)

Figure 2.

(a) The layout of the MC clusters embedded in both breast phantoms. Eight MC clusters, with a size range of 180–200, 250–280, 200–212, 280–300, 212–224, 300–355, 224–250, and 355–425 μm , respectively, were evenly spaced along a 2.8 cm radius circle. Each cluster consisted of 5×5 MCs of the same size range. (b) A MC image of the CT slices of large breast phantom.

Figure 3.

An example of six sequential MC slices of the small phantom imaged at 80 kVp and 12 mGy. Several but not all MC clusters were visible from 7th to 12th CT slices because it is very difficult to image MCs at the same CT slice while the embedding MCs in the phantoms and positioning the phantom for imaging. Each image size is 800×800 in pixel.

Figure 4 (a)

Figure 4 (b)

Figure 4.

The percentage of the visible MCs was plotted as a function of the MGD for eight MC size groups for (a) the small and (b) the large phantom imaged at 60 kVp, 80 kVp, and 100 kVp.

Figure 5.

The normalized noise level was plotted as a function of the MGD for the small and the large phantoms imaged at different x-ray tube voltages.

Figure 6 (a)

Figure 6 (b)

Figure 6.

The percentage of the visible MCs was plotted as a function of the x-ray tube voltage for eight MC size groups for (a) the small phantom at 3, 6, and 12 mGy, and (b) the large phantom at 6, 12, and 24 mGy.

Figure 7(a)

Figure 7(b)

Figure 7.

The percentage of the visible MCs was plotted as a function of the average MC size for (a) the small and (b) the large phantoms imaged at 80 kVp and three MGD levels. The data points for each combination of the MGD and the kVp were fit by Gaussian integral function for both breast phantoms.

Table 1

Imaging techniques for image acquisition.

Breast Phantom	MGD	X-ray tube voltage		
		60 kVp	80 kVp	100 kVp
Small	3 mGy	6 mA, 30 msec	7 mA, 10 msec	—
	6 mGy	13 mA, 30 msec	17 mA, 10 msec	8 mA, 10 msec
	12 mGy	26 mA, 30 msec	12 mA, 30 msec	21 mA, 10 msec
Large	6 mGy	16 mA, 30 msec	20 mA, 10 msec	9.2 mA, 10 msec
	12 mGy	33 mA, 30 msec	13 mA, 30 msec	6.3 mA, 30 msec
	24 mGy	—	26 mA, 30 msec	13 mA, 30 msec

Table 2

Parameters for image acquisition.

X-ray tube voltage	60, 80 and 100 kVp
Anode/filter	W/AI
Focal spot size	0.6 mm
Flat panel detector	Varian Paxscan 4030CB
Pixel size and acquisition mode	194 μm (1 $\times$ 1, non-binning mode)
Data acquisition rate	7.5 frames per second
Scanning time	40 sec per 360° rotation
Number of projection views	300 per 360° rotation
Voxel size	145 μm
Mean glandular dose	10.5-cm: 3, 6 and 12 mGy
	14.5-cm: 6, 12 and 24 mGy
Image reconstruction algorithm	Feldkamp algorithm with pure ramp filter

Table 3

The percentage visibilities for various conditions for the small and the large breast phantoms. The numbers in parentheses are the standard errors.

Breast Phantom	MGD	X-ray tube voltage		
		60 kVp	80 kVp	100 kVp
Small	3 mGy	18.6 (3.4)	18.0 (3.2)	—
	6 mGy	58.0 (5.3)	59.8 (5.5)	45.0 (5.8)
	12 mGy	88.8 (3.6)	85.5 (4.3)	80.8 (4.0)
Large	6 mGy	14.7 (2.5)	16.6 (2.4)	15.2 (2.5)
	12 mGy	50.2 (5.0)	42.8 (4.7)	32.2 (4.4)
	24 mGy	—	69.9 (3.9)	65.5 (5.1)

the numbers are expressed in %.

P-values for comparison of MC visibilities for two different x-ray tube voltages for the small and large breast phantoms.

Table 4

Breast Phantom	X-ray tube voltage	Mean Glandular Dose			
		3 mGy	6 mGy	12 mGy	24 mGy
Small	60 kVp vs. 80 kVp	0.86	0.65	0.27	
	60 kVp vs. 100 kVp	—	0.017	0.021	
	80 kVp vs. 100 kVp	—	< 0.01	0.21	
Large	60 kVp vs. 80 kVp		0.33	0.056	—
	60 kVp vs. 100 kVp		0.79	< 0.01	—
	80 kVp vs. 100 kVp		0.57	< 0.01	0.25

Table 5
The minimum detectable MC sizes for various conditions for the small and the large breast phantoms.

Threshold	Breast Phantom	MGD	X-ray tube voltage		
			60 kVp	80 kVp	100 kVp
50%	Small	3 mGy	346	349	—
		6 mGy	282	280	301
		12 mGy	253	258	259
	Large	6 mGy	357	353	354
		12 mGy	296	304	320
		24 mGy	—	270	279
75%	Small	3 mGy	368	366	—
		6 mGy	311	308	329
		12 mGy	261	264	271
	Large	6 mGy	382	373	376
		12 mGy	325	333	343
		24 mGy	—	298	301