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Reduction in x-ray scatter and radiation dose for volume-of-interest (VOI) cone beam breast CT—a phantom study

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Abstract

With volume-of-interest (VOI) cone beam computed tomography (CBCT) imaging, one set of projection images are acquired with the VOI collimator at a regular or high exposure level and the second set of projection images are acquired without the collimator at a reduced exposure level. The high exposure VOI scan data inside the VOI and the low exposure full field scan data outside the VOI are then combined together to generate composite projection images for image reconstruction. To investigate and quantify scatter reduction, dose saving, and image quality improvement in VOI CBCT imaging, a flat panel detector based bench-top experimental CBCT system was built to measure the dose, the scatter-to-primary ratio (SPR), the image contrast, noise level, the contrast-to-noise ratio (CNR), and the figure of merits (FOMs) in the CBCT reconstructed images for two polycarbonate cylinders simulating the small and the large phantoms. The results showed that, compared to the FF CBCT technique, radiation doses for the VOI CBCT technique were reduced by a factor of 1.20 and 1.36 for the small and the large phantoms at the phantom center, respectively, and from 2.7 to 3.0 on the edge of the phantom, respectively. Inside the VOI, the SPRs were substantially reduced by a factor of 6.6 and 10.3 for the small and the large phantoms, the contrast signals were improved by a factor of 1.35 and 1.8, and the noise levels were increased by a factor of 1.27 and 1.6, respectively. As the result, the CNRs were improved by a factor of 1.06 and 1.13 for the small and the large phantoms and the FOMs were improved by a factor of 1.4 and 1.7, respectively.

Keywords

cone-beam computed tomography; volume-of-interest; scatter-to-primary ratio; radiation dose; contrast-to-noise ratio; figure of merit

1. Introduction

Mammography is currently the major tool for breast cancer screening and diagnosis. However, the effectiveness of mammography is limited by the fundamental problems of projection imaging, including scattered radiation and anatomical structure noise. As the result, the

detection of breast cancer may be comprised by false negative and false positive mammograms [Lewin *et al* 2002, Huynh *et al* 1998]. One potential solution to overcome the problem of overlapping cancers with breast anatomy is to acquire and display three dimensional (3D) images, which provide better visualization of breast tissues and lesions. A number of studies have been conducted to investigate the feasibility of cone beam computed tomography (CBCT) techniques for breast cancer screening and diagnosis, and the results showed that CBCT breast imaging significantly improves the detectability of small lesions [Boone *et al* 2001, Ning *et al* 2001, Chen *et al* 2002, Glick *et al* 2002, Boone *et al* 2004, Gong *et al* 2004, Ning *et al* 2004a, Boone *et al* 2005, Gong *et al* 2005a, Gong *et al* 2005b, Ning *et al* 2005].

Despite its simple design and promising results, radiation dose is a major concern for CBCT breast imaging. It is desirable to keep the breast dose for CBCT within the mean glandular dose (MGD) limit of 6 mGy [ACR 1999] for screening mammography, however, higher exposure level may be required for imaging smaller structures such as microcalcifications. Therefore, volume-of-interest (VOI) imaging techniques have been developed and used to obtain equal or better image quality while applying sufficiently high exposures to the breast inside the VOI but reducing the overall breast dose [Chen *et al* 2008]. VOI techniques can be achieved by two approaches. One is to use a VOI filter, which has an opening to deliver x-ray exposure to a selected region containing the object of interest while attenuating direct x-ray exposure to regions outside the VOI. With this approach, the VOI filter is placed between the x-ray source and the patient and one full rotation scan is performed to generate high exposure projection image data inside the VOI but reduced exposure projection image data outside the VOI. Chityala *et al* [2004,2005] have used this approach to implement their region-of-interest (ROI) CBCT technique for high resolution angiography and the reconstructed images of a human head phantom were used to demonstrate its advantages. They conducted a simulation study to obtain an image quality (of the reconstructed images) comparable to that of the full field CBCT images with the exposure level outside the ROI reduced to as low as 1/16 of the original exposure level. In addition, they claimed that the dose level outside the ROI was reduced by a factor of 7. However, this claim appeared to be based on indirect estimation from exposure reduction outside the VOI since no Monte Carlo simulation or physical measurement were mentioned or described.

The alternative approach is to use a VOI collimator, which also has an opening to deliver x-ray exposure to the selected region but blocks unnecessary x-rays to regions outside the VOI. As a result, patient dose may be reduced because of the limited x-ray field size and reduced x-ray scatter [Siewerdsen *et al* 2001]. Similar to the VOI filter approach, a VOI collimator is placed between the x-ray source and the patient but two full rotation scans are taken: one is acquired with the VOI collimator at the same exposure level as the full field scan (referred to as VOI scan) and the other is acquired without the collimator at a low exposure level (referred to as low exposure full field scan). The use of the VOI scan data only may cause truncated image data outside the VOI, and reconstruction directly from truncated VOI projection data introduces artifacts into the reconstructed images, such as incorrect CT numbers and bright boundary rings. To minimize artifacts in the reconstruction, low exposure full field (FF_{LE}) projection data can be used to fill in the truncated space. The projection image data from these two scans can then be combined together to generate composite projection image data: high exposure VOI projection image data inside the VOI and FF_{LE} projection image data outside the VOI, which functions similarly to the VOI filter approach. Chen *et al* [2008] used this approach to perform a Monte Carlo simulation based on the Geant4 package for the VOI CBCT technique as applied to breast imaging and reported qualitatively that the radiation doses and the scatter signals were substantially reduced both inside and outside the VOI with improved CNR inside the VOI.

In this study, scatter reduction, dose saving, and image quality improvement of the VOI CBCT technique measured on the basis of simulated breast phantoms and compared with those of the full field (FF) CBCT technique. The scatter reduction was evaluated using the scatter-to-primary ratio (SPR) and the image quality improvement using the contrast-to-noise ratio (CNR) and the figure of merit.

2. Materials and Methods

2.1 Principle for the VOI CBCT technique

With the VOI collimator approach to the VOI CBCT techniques, two full rotation scans are required: one is acquired with the VOI collimator at a high exposure level and the other without at a low exposure level. An exposure reduction ratio, α , can be defined as:

$$\alpha = X_{FF}^{LE} / X_{FF} \quad (1)$$

where X_{FF}^{LE} and X_{FF} are the total exposures measured at the isocenter for the FF_{LE} and the FF scans, respectively, with the FF scan without phantom present. For exposure reduction, α ranges from 0 to 1. To combine projection data from these two scans for VOI reconstruction, the VOI projection data (I_V) outside the VOI are replaced by the normalized FF_{LE} projection data (I_{FF}^{LE} / α) to generate composite projection images (I_C) whose signals are defined as follows:

$$I_C = \begin{cases} I_V, & \text{inside the VOI} \\ I_{FF}^{LE} / \alpha, & \text{outside the VOI} \end{cases} \quad (2)$$

where I_{FF}^{LE} is the image signals of the FF_{LE} projection data and normalized by the factor, $1/\alpha$, to compensate for exposure reduction outside the VOI. Notice that although I_{FF}^{LE} / α has the same average signal magnitude as I_{FF} , I_{FF}^{LE} / α has higher noise level than I_{FF} due to I_{FF}^{LE} used for acquisition. The composite projection images can be reconstructed in a similar manner as the FF projection images.

2.2 SPR estimation

X-ray scatter poses a significant challenge to CBCT imaging, resulting in reduced soft tissue contrast and increased artifacts in the reconstructed CBCT images [Siewerdsen *et al* 2001, Endo *et al* 2001, Kwan *et al* 2005]. Several scatter control techniques have been investigated to minimize the effects of x-ray scatter, including various scatter correction techniques [Spies *et al* 2001, Ning *et al* 2004, Liu *et al* 2006], scatter rejection techniques [Chen *et al* 2008, Endo *et al* 2001, Siewerdsen *et al* 2004]. The use of collimated x-rays in VOI scans helps reduce x-ray scatter. Since x-ray image signals consist of a primary component and a scatter component, Eq. (2) can be rewritten as follows:

$$I_C = \begin{cases} P_V + S_V, & \text{inside the VOI} \\ P_{FF}^{LE} / \alpha + S_{FF}^{LE} / \alpha, & \text{outside the VOI} \end{cases} \quad (3)$$

where P_V and S_V are the primary and the scatter signals from the VOI scan, and P_{FF}^{LE} and S_{FF}^{LE} from the FF_{LE} scan, respectively. Then, the SPR, defined as the ratio of the scatter signals to the primary signals, in the composite images may be expressed as follows:

$$SPR_c = \begin{cases} S_v/P_v = SPR_v, & \text{inside the VOI} \\ (S_{FF}^{LE}/\alpha)/(P_{FF}^{LE}/\alpha) = S_{FF}^{LE}/P_{FF}^{LE} = SPR_{FF}^{LE}, & \text{outside the VOI} \end{cases} \quad (4)$$

where SPR_v and SPR_{FF}^{LE} are the SPRs with the VOI and the FF_{LE} scans, respectively. However, since in terms of signal size, $P_{FF}^{LE}/\alpha = P_{FF}$ and $S_{FF}^{LE}/\alpha = S_{FF}$ on the basis of Eq. (1), therefore,

$$SPR_c = \begin{cases} SPR_v, & \text{inside the VOI} \\ SPR_{FF}, & \text{outside the VOI} \end{cases} \quad (5)$$

Notice that SPR_v is substantially lower than SPR_{FF} due to the use of x-ray collimation during acquisition. Notice also that the objective of using the VOI CBCT technique is to obtain high quality reconstructed images inside the VOI. Thus, the reduction of SPR inside the VOI would in principle help improve the image quality there.

2.3 CBCT imaging system

A bench-top experimental flat-panel (FP) based cone beam breast CT system was constructed and used to image phantoms and mastectomy breast specimens [Lai *et al* 2007, Yang *et al* 2007, Altunbas *et al* 2007]. The system consisted of an amorphous silicon/cesium iodide (a-Si:H/CsI) FP detector (Paxscan 4030CB, Varian Medical Systems, Salt Lake City, UT), a conventional tungsten target x-ray tube (G-1592/B180, Varian Medical Systems, Salt Lake City, UT) with nominal focal spot sizes of 0.6 mm (used in this study) and 1.2 mm, and a rotating table for holding and rotating breast specimens and simulated phantoms (Fig. 1). The x-ray source-to-isocenter distance (SCD) and the x-ray source-to-image detector distance (SID) were 75 cm and 104 cm, respectively. In order to simulate pendant geometry cone beam breast CT, a 3 mm thick lead shield was directly mounted in front of the x-ray tube to block half of the x-ray beam, resulting in a half-cone x-ray beam for scanning the breast. Images were acquired at 7.5 frames/sec in the non-binning mode (i.e. pixel size = 194 μ m) with exposures made at 80 kVp.

2.4 Breast phantoms

To investigate the effect of breast size on the SPR, a 11-cm diameter polycarbonate (lexan) cylinder and a 15-cm diameter one were constructed and used to simulate small and large breasts, respectively. Both phantoms consisted of three layers, two 9.5 cm in height and one 2.5 cm in height. One 9.5-cm thick layer was solid and the other had a through-hole at the center of the cylinder. The layers were stacked together to hold an ion chamber for exposure measurements (Fig. 2, top). The 2.5-cm layer with through-holes drilled at locations at various distances from the center may be placed on top of the 9.5-cm thick solid layer and used to hold thermoluminescent dosimeters (TLDs) for dose measurements or iodine contrast solutions for CNR measurement (Fig. 2, bottom). The phantoms were centered with the rotating axis and aligned with the x-ray beam for measurements. A 25.4-cm diameter, 5 cm high polycarbonate cylinder was placed on top of the breast phantom to mimic the chest wall (Fig. 2, bottom).

2.5 VOI collimator

To simulate VOI cone beam breast CT, a 2.25 mm thick lead with a rectangular opening was used as the VOI collimator. The VOI collimator was placed between the x-ray tube and the breast phantom at 39 cm from the x-ray focal spot to cover a 2.5 $\times$ 2.5 cm region at the isocenter. The center of the region was aligned with the rotating axis to define a cylindrical VOI centered

with the rotating axis. The top edges of the VOI opening and the x-ray beam were aligned with each other (Fig. 2, bottom). The VOI collimator was kept stationary during the scans.

2.6 SPR measurement

The SPRs were measured in the projection images only for a single view due to the symmetry of the scanning geometry. To study the horizontal variation of the SPR across the detector, a horizontal slot collimator made of 4 mm thick lead was placed between the x-ray tube and the breast phantom to estimate the primary signals along the slot. The top edge of the slot was aligned with the top edge of the x-ray beam. The slot width was adjustable in the axial direction (z-axis) in order to change the FOV. Image signals within the slot consisted of primary signals and scatter components that decrease with the slot width due to reduced FOV. The slot width was varied from 3.8 to 12.8 mm with an increment of 1.8 mm. For each slot width, twenty projection images were acquired and averaged to minimize the signal fluctuations. Then, the phantoms and the collimator were removed and additional twenty projection images were acquired, averaged and used as the reference image. The x-ray tube was operated at high and low mAs settings for the FF and the FF_{LE} scans, respectively. The exposures were measured by an ion chamber (10×5-6, Radical Corp, Monrovia, CA) at the isocenter without phantoms present. The ratio of the low exposure to the high exposure was calculated based on the definition in Eq. (1) and was about 0.258. The SPR measurements were first performed to calculate the normalized image signal (NIS), defined as the total signals from the slot image divided by those from the reference image, for each slot width. Then, the primary signals were determined by extrapolating the linear fit between the NIS and the slot width to zero width. The scatter signals were computed as the difference between the total signals and the primary signals as the primary signals do not change with the FOV or the slot width. The spatial distribution of the SPR at the detector plane was then computed for the FF, the FF_{LE}, and the VOI projection scans. The total signals, primary signals, scatter signals, and the SPRs were then transferred from the detector plane to the isocenter plane. Total, primary, and scatter signals from the VOI and the FF_{LE} scans were used to compute the spatial variations of the total signal, the primary signal, the scatter signal, and the SPR for the composite images.

2.7 CNR measurement

To investigate the effect of the VOI CBCT technique on image contrast, iodine solutions of two different concentrations were used to fill the through-holes in the phantoms: 2.91 and 4.34 mg/ml for the small phantom and 4.34 and 8.55 mg/ml for the large phantom. The contrast signal was defined and measured as the difference in the CT number between the average voxel value in the through-hole regions (filled with iodine solutions) and that in the surrounding area (lexan). Because the visibility of an object is determined not only by the contrast of the object but also by the voxel noise in a tomographic image, CNRs in the CBCT images were computed and used to evaluate the image quality. The voxel noise level was defined and measured as the standard deviation of the image signals in the surrounding background regions. For each full rotation scan, 300 projection views were acquired, with an angular increment of 1.2° over 360°, while the x-ray tube was operated in pulsed mode at 7.5 pulses/sec. The x-ray techniques for image acquisition with the FF and the VOI rotation scans were 372 mAs and 427 mAs for the small and the large phantoms, respectively, and 110 mAs and 128 mAs, respectively, with the FF_{LE} rotation scan. Image reconstruction was performed using Feldkamp's backprojection (FDK) algorithm [Feldkamp *et al* 1984] with a pure ramp filter on a PC cluster (24 PCs with dual Intel Xeon processors (2.4 GHz)) for the FF CBCT images based on the FF projection image data and the composite projection data for the VOI CBCT images. The CNRs were then computed at various through-hole locations for the FF CBCT and the VOI CBCT images and plotted as a function of the radial distance from the center of the phantom.

2.8 Dose measurement

To measure the doses with the TLDs, a calibration measurement was first made at the center of the phantom with the FF full rotation scan using an ion chamber (10×5-6, Radical Corp). The active volume of the ion chamber was positioned right beneath the rotating plane or the top of the half cone beam (top, Fig. 2). The linearity between the exposure (unit in R) and current-exposing time (in mAs) for pulsed x-rays was evaluated and found to be good ($r^2 = 0.994$). Different combinations of the tube current and the pulse width (either 14 or 33 msec) were used to achieve various exposure levels. Exposures were then converted to absorbed dose in tissue using the conversion factor 0.873 (rad/R) and the tissue-air ratio of mass energy absorption coefficient $(\mu_{ab}/\rho)_{air}^{tissue}$. For dose measurements, TLD capsules were placed in the through-hole at the phantom center of the disk (Fig. 2, bottom) and exposed with the same techniques used in the exposure measurements. Thermoluminescent signals were then read out using a TLD reader (Harshaw 3500 TLD reader, Harshaw-Bicron, Newbury, OH) and plotted as a function of absorbed dose obtained from the exposure measurement as the calibration curve. To study dose variation with locations, TLD capsules were placed in the through-holes at various locations of the phantom and exposed to pulsed x-rays with the same imaging techniques as those used for the CNR measurements. The absorbed doses were then computed using the calibration curve and plotted as a function of the radial distance from the phantom center for the FF and the VOI scans. The doses for the FF_{LE} rotation scan were calculated by multiplying the exposure reduction ratio (α) with the doses for the FF scan. The doses for the VOI CBCT technique were computed by adding the doses from the VOI and the FF_{LE} scans.

3. Results

3.1 Scatter reduction

Figures 3(a)-(c) show the measured primary, scatter, and the total signals, respectively, along a horizontal line with the FF, the VOI, and the FF_{LE} scan images for the small phantom. As observed, with the FF and the FF_{LE} scan images, the primary, the scatter, and the total signals increased with the radial distance from the center of the phantom but with different signal magnitudes because of exposure reduction ratio (α). With the VOI scan image, both primary and scatter signals outside the VOI were substantially reduced as compared to those with both FF and FF_{LE} scan images due to x-ray attenuation by the VOI collimator. The primary signals inside the VOI were similar to those with the FF scan image because the same incident exposure level was used for both VOI and FF scans. The scatter signals inside the VOI were substantially reduced due to a smaller FOV. As the result, the total signals with the VOI scan image were lower than those with the FF scan image inside the VOI. The composite images were formed with the image signals in the VOI image inside the VOI and those in the FF image outside the VOI. Similar findings were observed for the large phantom. Figures 4(a) and (b) show the SPR plotted as a function of the radial position with the FF and the FF_{LE} scan images, and the composite image for the small and the large phantoms, respectively. It was observed that the FF_{LE} image had similar SPRs to the FF image but with more fluctuations for both phantoms. With both FF and FF_{LE} images, the SPRs peaked at around the center of the phantom and decreased gradually toward the edge of the phantom. Figure 4 also shows that the composite image inherited the SPRs from the FF_{LE} image outside the VOI and from the VOI image inside the VOI. Reduced SPRs inside the VOI were observed, which is expected due to substantially reduced scatter (fig. 3(b)). The SPR reduction factors, defined as the ratio of the SPR averaged over 50 pixels at the center of the VOI for the FF scan image to those for the composite image, were computed and found to be $6.6 (\pm 0.8)$ and $10.3 (\pm 1.7)$ for the small and the large phantoms, respectively.

3.2 Contrast signals, noise, and CNR

Figure 5 shows an example of the measured contrast signals of 4.34 mg/ml iodine solution plotted as a function of the radial position with the FF and the VOI CBCT images for both phantoms. It was observed that the VOI CBCT images had significantly higher contrast signals than the FF CBCT images inside the VOI but similar contrast signals outside the VOI. The contrast improvement factors (CIFs), defined as the ratios of the contrast signals for the VOI CBCT images to those for the FF CBCT images, were computed at the center of the phantom. The CIFs for the two contrast levels were found to be similar and further averaged. The mean CIFs were found to be $1.35 (\pm 0.09)$ and $1.8 (\pm 0.08)$ for the small and the large phantoms, respectively. In Figure 6, the voxel noise levels were computed and plotted as a function of the radial position with both CBCT images and both phantoms. The results showed that the VOI CBCT images had higher noise levels than the FF CBCT images inside and outside the VOI for both phantoms. The ratios of the noise levels with the VOI CBCT images to those with the FF CBCT images were computed inside the VOI and found to be $1.27 (\pm 0.07)$ and $1.6 (\pm 0.05)$ for the small and the large phantoms, respectively.

The CNRs were plotted as a function of the radial position with both CBCT images and both phantoms in Figure 7. The plots showed that the VOI CBCT images had comparable CNRs to the FF CBCT images inside the VOI for both phantoms. The CNR improvement factors (CNRIFs), defined as the ratios of the CNR for the VOI CBCT images to those for the FF CBCT images, were computed at the center of the phantom. The CNRIFs for both contrast levels were found to be similar and then averaged. The mean CNRIFs were found to be $1.06 (\pm 0.05)$ and $1.13 (\pm 0.07)$ for the small and the large phantoms, respectively.

Figures 8 and 9 show the phantom layout and the axial images for the FF CBCT and the VOI CBCT images for the small and the large phantoms, respectively. The images were printed with the same window and level settings for comparison. Five circular objects were observed in the small phantom with both CBCT axial images, and six were observed in the large phantom. However, the edges of the objects outside the VOI in the VOI CBCT images appeared to be fussier than those in the FF CBCT images, indicating higher noise level for the VOI CBCT images as compared to that for the FF CBCT image due to the low transmission of x-ray fluence. The circular objects inside the VOI in the VOI CBCT images appeared to have higher contrast than those with the FF CBCT images.

3.3 Dose reduction

Figures 10(a) and (b) show the measured dose plotted as a function of the radial position for the FF and the VOI scans for the small and the large phantoms, respectively. The radiation doses for the VOI CBCT technique were estimated from the addition of the VOI scan and the FF scan but multiplying with the exposure reduction ratio (α) because of the FF_{LE} scan. Figure 10 shows that the doses with the VOI CBCT technique ($\alpha=0.258$ in this study) were substantially reduced at all locations (both inside and outside the VOI) as compared to those with the FF CBCT technique. The dose reduction factors, defined as the ratios of the doses with the FF CBCT technique to those with the VOI CBCT technique, were computed at all locations and found to increase with the radial position: $1.20 (\pm 0.13)$ and $1.36 (\pm 0.17)$ for the small and the large phantoms, respectively, at the phantom center, and $2.7 (\pm 0.3)$ and $3.0 (\pm 0.4)$, respectively, at the edge of the phantom.

3.4 Figure of merit

To normalize image quality to the radiation risk to the patient, the figure of merits (FOMs), defined as $CNR^2/dose$, were computed for both FF CBCT and VOI CBCT images and both phantoms. Figure 11 shows an example of the FOMs plotted as a function of the radial position for 4.34 mg/ml iodine solution. The plots showed that the VOI CBCT images had similar or

higher FOMs than the FF CBCT images inside the VOI but lower FOMs outside the VOI. The FOM improvement factors (FOMIFs), defined as the ratios of the FOMs for the VOI CBCT images to those for the FF CBCT images, were computed and averaged for the two different contrast levels at the center of the phantom. They were found to be $1.4 (\pm 0.2)$ and $1.7 (\pm 0.3)$ for the small and the large phantoms, respectively.

4. Discussion

There are several key observations in this study. The SPRs inside the VOI were substantially reduced in the composite images used with the VOI CBCT technique (Figure 4). This is because VOI scan data were used there and they were acquired with x-rays collimated by the VOI mask. As the result, the contrast signals in the VOI CBCT images were significantly improved as compared to the FF CBCT images (Figure 5). However, no significant difference in image contrast was observed between the FF CBCT and the VOI CBCT images outside the VOI. This is because the composite projection images outside the VOI were filled with the normalized FF_{LE} projection data, which had nearly identical SPRs as the FF projection data. As the result, the FF CBCT and the VOI CBCT images had similar image contrast. However, in regions right outside the VOI, the VOI CBCT images had higher contrast signals than the FF CBCT images. This may be due to the fact that the contrast signals were measured on reconstructed images which were formed with various combinations of VOI scan data and normalized FF_{LE} scan data depending on the location of the voxels. For each voxel, projection data corresponding to all x-ray paths going through the voxel and covering 360° view angles were backprojected and summed up to form the reconstructed image data. For voxels near the VOI, more rays went through the VOI, thus more VOI scan data, which has substantially lower SPRs, were used for the reconstruction. Since the VOI scan data have considerably smaller scatter component, the reconstruction was subject to smaller scatter effect thus resulting a higher contrast signal. As the voxels moved away from the VOI, fewer VOI scan data and more normalized FF_{LE} scan data, which have the same high SPRs as the FF scan data, were used for reconstruction and thus the contrast degradation effect is mainly determined by the high SPRs of the FF_{LE} scan data. Another finding is that the contrast signals inside the VOI decreased with the radial position, which appeared to be statistically insignificant (as indicated by the error bar). These decreases appear to be consistent for both phantoms and both FF CBCT and VOI CBCT images. This phenomenon may be related to the fact that the SPR varies little among various view angles at the phantom center while varying significantly near the boundary of the VOI opening, which was also observed that image noise inside the VOI increased with the radial position. However, a more detailed study of these phenomena is beyond the scope of this study.

Increased noise was expected inside and outside the VOI with the VOI CBCT images (Fig. 6). This can be explained by the fact that the image signals with the VOI scan were reduced inside the VOI due to less scatter. Thus, the signal-to-noise ratios (SNRs) of the VOI projection images were expected to be lower than those of the FF projection images. Outside the VOI, the composite projection data were filled with the FF_{LE}/α projection data, whose SNRs were also lower than the FF projection images due to the low exposure level. Therefore, the composite projection images had lower SNRs than the FF projection images. Since CBCT images are reconstructed from logarithmically mapped projection image data, the VOI CBCT images had a higher signal level than the FF CBCT images. As a result, the noise level was increased, which was observed in Figs. 8 and 9: the VOI CBCT images look noisier than the FF CBCT images, especially for the areas outside the VOI (the yellow circle). No significant difference in image noise between the FF CBCT and the VOI CBCT images was visualized inside the VOI because the primary x-ray signals still remained the same inside the VOI even though x-ray scatter was significantly reduced for the VOI images, so that the signals of the composite images were close to those with the FF projection images (Fig. 3(a)). The results also showed that the VOI CBCT images had slightly increased CNRs (CNRIF = 1.06 and 1.13 for the small

and the large phantoms, respectively, at the phantom center) but significantly improved FOM (FOMIF = 1.4 and 1.7 for the small and the large phantoms, respectively). This is due to the trade-off between the increased image contrast and the increased noise level in the reconstructed images from reduced x-ray scatter. Eqs. (3) and (6) imply that the SPR and image noise with the VOI images are independent of α , indicating that the CNRs should remain the same. However, the FOM may vary depending on the α value due to radiation dose. Figures 12 (a) and (b) show dose reduction factor plotted as a function of $1/\alpha$ at the center and on the edge of the phantom, respectively. Both figures show that the dose reduction factor increases with the $1/\alpha$. At the phantom center, the dose reduction factor ranges from 0.64 (when $\alpha=1$) to 1.75 (when $\alpha=0$) for the small phantom and from 0.68 to 2.1 for the large phantom; and from 0.9 to 9.3 for the small phantom and from 0.93 to 12.6 for the large phantom on the edge of the phantom. Notice that there was no dose reduction at the phantom center as α larger than 0.42 and 0.52 for the small and the large phantoms, respectively. The results indicate that radiation dose can be reduced by a factor of up to 2 at the phantom center and by a factor of 10 on the edge of the phantom for the VOI CBCT technique and imply that there is a possibility to reduce noise level and to improve low contrast performance inside the VOI with higher exposure to the VOI while keeping dose to the rest of the breast as low as possible. Chen et al [2008] demonstrated and reported that α can be as low as 0.1 for the FF_{LE} projection images with improved CNR for the VOI CBCT images. The results also showed that improvement with the VOI CBCT technique varies with the phantom size: the large phantom had the higher values in the SPR reduction factor (10.3 vs. 6.6), the CIF (1.8 vs. 1.35), the CNRIF (1.13 vs. 1.06), and the FOMIF (1.7 vs. 1.4) than the small phantom, indicating that the VOI CBCT technique may be more beneficial for the large phantom. This may be because degree of the scatter reduction was more pronounced for the large phantom and so were the improvement of the CNRs and the FOMs.

In forming the composite projection images for VOI CBCT, the VOI scan data were used for the region inside the VOI while the FF_{LE} scan data were used for outside the VOI. The composite images may be alternatively formed by adding the FF_{LE} scan data to the VOI scan data for inside the VOI. This seems to have the advantage of using all primary x-rays available. We have measured the SPR, contrast signals, and CNRs for this approach. The results show that there was actually a decrease of the CNRs as compared to the approach using the VOI scan data alone and even the FF CBCT technique. This may be because a considerable amount of scatter was included with the FF_{LE} scan data added. This scatter component was even greater than that in the VOI scan data and caused the noise level to increase by a factor of greater than $\sqrt{2}$ while increasing the primary signals only by about 26%. Thus, we concluded that the approach using the VOI scan data only (for inside the VOI) is still the better approach.

There were several limitations for this study. First, all of the measurements were performed close to the focal spot plane. Further investigations at regions away from the focal spot center (along the z-direction) are needed. Second, the center of the VOI collimator is stationary at the isocenter during the scans based on the assumption that the suspicious lesion is located at the center of the breast. However, if the VOI is located away from the center of the breast, the VOI shall rotate around the isocenter during the scan. Thus, a moving VOI collimator needs to be used to track and acquire images of the off-center VOI. A scan system with such capability is being developed and tested in our lab and will be reported in the near future. Thirdly, since the FDK algorithm was used to reconstruct the CBCT images, the computing time for the VOI CBCT technique is only slightly longer than that for the FF CBCT technique. The addition time is required to pre-process the projection images by combining the VOI scan data with the normalized FF_{LE} scan data. Currently, the pre-processing and reconstruction times based on the use of a 64-CPU cluster were about 5 and 25 minutes for 300 projection views and 800 slices of 800×800 images, respectively. However, the computing time will likely be

substantially shortened with further optimization or the use of graphic processing unit (GPU) both of which are being developed and investigated in our lab.

5. Conclusion

We investigated and measured the SPR, the radiation dose, the CNR, and the FOM with the VOI CBCT technique employing the VOI collimator, which can achieve significantly reduced dose inside and outside the VOI in addition to reduced SPR inside the VOI. Compared to the FF CBCT images, improved contrast signal, CNR, and FOM inside the VOI were found for the VOI CBCT images. This study demonstrated that the VOI CBCT technique can be used to image a pre-selected region to improve image quality inside the VOI with reduced breast dose.

General summary

The purpose of this study is to compare volume-of-interest (VOI) cone beam computed tomography (CBCT) with full field (FF) CBCT techniques by physical measurements. Two cylindrical polycarbonate phantoms with different dimensions used to simulate breasts were imaged with an experimental CBCT system at 80 kVp. The FF scan was performed at high exposure level (HEL). Two scans were performed with the VOI CBCT technique: one with FF at low exposure level (LEL) and the other with the VOI collimator present using the same technique for the HEL. The ratio of the LEL to the HEL was 0.258. Scatter-to-primary ratio (SPR) and radiation dose were measured, and image quality was quantified by measuring the contrast-to-noise ratio (CNR) and figure of merit (FOM). The results showed that with the VOI CBCT technique, the SPR and the radiation dose were substantially reduced, the CNR was slightly improved, and the FOM was significantly improved.

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Figure 1.
Experimental cone beam computed tomography imaging system.

Figure 2.

The layout of the two identical cylinders stacked together for the exposure measurement (top). The active volume of the ion chamber was aligned with the x-ray beam. The layout of the 2.5-cm disk stacked between the two cylinders for the SPR, dose, and the CNR measurements (bottom). The TLDs and the iodine solutions were aligned with the x-ray beam.

Figure 3.

- (a). Primary signal at the isocenter plane versus radial position for the small phantom with the FF, the VOI, and the FF_{LE} scan images.
- (b). Scatter signal at the isocenter plane versus radial position for the small phantom with the FF, the VOI, and the FF_{LE} scan image.
- (c). Total signal at the isocenter plane versus radial position for the small phantom with the FF, the VOI, and the FF_{LE} scan image.

Figure 4.

- (a). The scatter-to-primary ratio at the isocenter plane versus radial position for the small phantom with the FF and the FF_{LE} scan images, and the composite image.
 (b). The scatter-to-primary ratio at the isocenter plane versus radial position for the large phantom with the FF and the FF_{LE} scan images, and the composite image.

Figure 5. Contrast signal (difference in CT number) versus radial position for the small and the large phantoms with the FF CBCT and the VOI CBCT images.

Figure 6. Noise level (difference in CT number) versus radial position for the small and the large phantoms with the FF CBCT and the VOI CBCT images.

Figure 7. Contrast-to-noise ratio versus radial position for the small and the large phantoms with the FF CBCT and the VOI CBCT images.

Figure 8.

The sketch of the through-holes of the small phantom (a) for the thermoluminescent dosimeters and iodine solutions. The phantom images were acquired with the FF CBCT (b) and the VOI CBCT (c) techniques. The image matrix for both images is 768×768 pixels with the same window/level settings. The VOI field (yellow circle) was about 2.5 cm in diameter at the isocenter.

Figure 9.

The sketch of the through-holes of the large phantom (a) for the thermoluminescent dosimeters and iodine solutions, and the phantom images were acquired with the FF CBCT (b) and the VOI CBCT (c) techniques. The image matrix for both images is 1044×1044 pixels with the same window/level settings. The VOI field (yellow circle) was about 2.5 cm in diameter at the isocenter.

Figure 10.

- (a). Radiation dose versus radial position for the small phantom imaged with the FF CBCT technique and the VOI CBCT technique with varying α .
 (b). Radiation dose versus radial position for the large phantom with the FF CBCT technique and the VOI CBCT technique with varying α .

Figure 11.

The figure of merit versus radial position for the small and the large phantoms with the FF CBCT and the VOI CBCT images.

Figure 12.

(a). The dose reduction factor at the center of the phantom versus $1/\alpha$ with the VOI CBCT technique for the small and the large phantoms.

(b). The dose reduction factor at the edge of the phantom versus $1/\alpha$ with the VOI CBCT technique for the small and the large phantoms.