Feasibility of volume-of-interest (VOI) scanning technique in cone beam breast CT—a preliminary study

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This work is to demonstrate that high quality cone beam CT images can be generated for a volume of interest (VOI) and to investigate the exposure reduction effect, dose saving, and scatter reduction with the VOI scanning technique. The VOI scanning technique involves inserting a filtering mask between the x-ray source and the breast during image acquisition. The mask has an opening to allow full x-ray exposure to be delivered to a preselected VOI and a lower, filtered exposure to the region outside the VOI. To investigate the effects of increased noise due to reduced exposure outside the VOI on the reconstructed VOI image, we directly extracted the projection data inside the VOI from the full-field projection data and added additional data to the projection outside the VOI to simulate the relative noise increase due to reduced exposure. The nonuniform reference images were simulated in an identical manner to normalize the projection images and measure the x-ray attenuation factor for the object. Regular Feldkamp–Davis–Kress filtered backprojection algorithm was used to reconstruct the 3D images. The noise level inside the VOI was evaluated and compared with that of the full-field higher exposure image. Calcifications phantom and low contrast phantom were imaged. Dose reduction was investigated by estimating the dose distribution in a cylindrical water phantom using Monte Carlo simulation based Geant4 package. Scatter reduction at the detector input was also studied. Our results show that with the exposure level reduced by the VOI mask, the dose levels were significantly reduced both inside and outside the VOI without compromising the accuracy of image reconstruction, allowing for the VOI to be imaged with more clarity and helping to reduce the breast dose. The contrast-to-noise ratio inside the VOI was improved. The VOI images were not adversely affected by noisier projection data outside the VOI. Scatter intensities at the detector input were also shown to decrease significantly both inside and outside the VOI in the projection images, indicating potential improvement of image quality inside the VOI and contribution to dose reduction both inside and outside the VOI. © 2008 American Association of Physicists in Medicine. [DOI: [10.1118/1.2948397](https://doi.org/10.1118/1.2948397)]

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I. INTRODUCTION

Breast cancer is recognized as a major threat to women's health. One in eight women in the United States may be diagnosed with breast cancer in her lifetime.¹ Early detection is critical to improved survival and decreased mortality in breast cancer patients.^{2–5} Currently, mammography is the primary method used for breast cancer screening. However, a major limitation of mammography is overlapping of breast tissue structures with soft tissue masses or microcalcifications in the two-dimensional (2D) mammograms. This compromises its use in the detection, diagnosis, and staging of breast cancer.⁶

We and other investigators are studying the feasibility of using a pendant geometry cone beam CT scanner for breast cancer imaging.^{7–9} With pendant geometry cone beam breast CT, only the breast itself is exposed to the x rays, preventing

unnecessary radiation exposure to other parts of the body and providing isotropic spatial resolution 3D images. However, in diagnostic workup or staging, it may not be necessary to achieve the same high image quality throughout the entire breast. The radiologist or surgeons may be interested in a small volume with high image quality. In this situation, it is desirable to image the volume of interest (VOI) and limit the exposure to the VOI only. However, reconstruction directly from truncated projection data introduces artifacts into reconstructed images,^{10,11} such as central cupping, incorrect CT numbers, and bright boundary rings. Although some researchers have been seeking alternative algorithms to allow image reconstruction directly from truncated or limited projection data,^{12–18} their algorithms either require special x-ray source trajectories, such as spiral trajectory,^{12–14} or limit number of truncated views.¹⁶ Pan *et al.*,¹⁵ Wang *et al.*,¹⁹

Chen *et al.*,²⁰ and Noo *et al.*²¹ developed a backprojection-filtration (BPF) algorithm for reconstructing truncated fan beam or cone beam projection data. Their method works only for projections truncated for a peripheral region of interest (ROI) on one side of the object, but is subject to artifacts when the ROI is inside of the object and results in truncation on both sides of the ROI. Although the data extrapolation approach proposed by Hsieh *et al.*¹⁶ can do reconstruction from projections with two-side truncations, it still requires completed projections in some projection views.

The Feldkamp–Davis–Kress (FDK) reconstruction of a fixed point of an object requires the knowledge of the projections away from the point at the same longitudinal position. To minimize these artifacts in the reconstruction of a VOI, the knowledge of projection data, outside the VOI and at the same longitudinal position, is required. Thus, to achieve good image quality, it is generally necessary to cover a large volume of the object with the x-ray field. To minimize the radiation dose to the object, x-ray exposure outside the VOI may be reduced by inserting a VOI mask, which collimates the unnecessary x-ray below and above the VOI and reduces the x-ray exposure within the VOI height, between the x-ray source and the object. Chityala *et al.*¹⁸ have used this approach to implement their ROI cone beam CT technique for neuro-angiography and Graham *et al.*²² investigated its application in radiation therapy. To our knowledge, however, this approach has not been studied in breast cancer imaging.

In this study, therefore, we investigated a VOI cone beam CT technique for 3D breast imaging. With this technique, x rays are attenuated outside the VOI to generate reduced exposure, noisier images but unattenuated inside the VOI to generate full exposure, less noisy images. The projection data are normalized by similarly acquired reference images to obtain attenuation data, which are then reconstructed with the regular FDK algorithm. Effects of exposure reduction outside the VOI on the reconstructed images inside the VOI were studied. Potential dose saving and scatter reduction abilities of the VOI technique were demonstrated by means of Monte Carlo simulation.

II. MATERIALS AND METHODS

II.A. Principle

The VOI scanning technique is designed to minimize the dose to the breast in cone beam breast CT during diagnostic and staging procedures without affecting the image quality in the VOI. This is achieved by inserting a specially designed mask between the x-ray source and the breast as shown in Fig. 1. The mask has a circular or rectangular opening to deliver a full exposure to a preselected spherical or cylindrical VOI and a lower level exposure to the region outside the VOI. During the scan, this mask may be continually moved up and down and left and right to point to the opening towards the VOI if the VOI is off center. Assuming that the x-rays used are mono energetic, the attenuated x-ray image

FIG. 1. Diagram for cone beam breast CT with VOI scanning technique.

signals inside and outside the opening of the VOI mask [Fig. 2(a)], I^{in} and I^{out} , may be expressed according to the Lambert–Beers law, as follows:

$$I^{\text{in}} = I_0 e^{-\sum_i \mu_i t_i}, \quad I^{\text{out}} = I_0 e^{-\mu_f t_f} e^{-\sum_i \mu_i t_i}, \quad (1)$$

where I_0 is the x-ray intensity emitted from x-ray source, μ_i and μ_f are linear attenuation coefficients of the breast tissue and the VOI mask; t_i and t_f are their corresponding thicknesses, respectively. To prepare for reconstruction, each projection image is normalized by a nonuniform reference image acquired with the VOI mask at the same location shown in Fig. 2(b). The reference signals inside and outside the opening of the VOI mask may respectively be expressed as follows:

$$I_0^{\text{in}} = I_0, \quad I_0^{\text{out}} = I_0 e^{-\mu_f t_f}. \quad (2)$$

Using these signals, the logarithmic attenuation data can be computed in the same manner inside and outside the opening of the VOI mask:

$$\sum_i \mu_i t_i = \ln \left(\frac{I_0^{\text{in}}}{I^{\text{in}}} \right) = \ln \left(\frac{I_0^{\text{out}}}{I^{\text{out}}} \right). \quad (3)$$

Notice that although the signals inside and outside the VOI differ by the factor, $e^{-\mu_f t_f}$, this factor cancels out in normalization with the nonuniform reference images. Thus, these attenuation data can be processed in a similar manner as regular full-field cone beam CT projection images prior to reconstruction. Alternatively, the projection images inside the opening of the VOI mask may be mapped to equalize the signal magnitude outside the VOI for reconstruction.¹⁷

II.B. Simulation of low exposure signals

We have constructed and used a bench top experimental cone beam CT system to image postmastectomy breast specimens. As shown in Fig. 3, the system consists of a radiographic x-ray tube (Varian Medical Systems, Salt Lake City, UT) with 0.6 mm nominal focal spot size, a Varian Paxscan 4030 CB flat panel detector, and a step motor driven rotating table to hold and rotate the object.²³ The x-ray source to detector and source to rotating center distances were 104 and 75 cm, respectively. Three hundred projection images were acquired for each 360° scan. The detector pixel size and reconstructed voxel size were 194 and 140 μm , respectively. The full-field cone beam CT images of postmastectomy

FIG. 2. Illustration of the VOI technique and corresponding x-ray signal intensities.

breast specimens were acquired at 80 kVp, with the exposure set to generate an iso-center dose $2\times$ of the mean glandular dose (MGD) limit (0.6 rads for two view mammograms).²⁴

To simulate the projection images acquired with the VOI mask, we computed signals and noises corresponding to the lower radiation exposure level on each projection data outside the VOI while leaving the projection data within the VOI unchanged. Our simulation method was to scale down the full exposure signals by a factor and then add a calculated amount of random noise to the scaled images. Changes in x-ray scatter and beam hardening due to attenuation and filtering by the VOI mask were ignored so that we could focus on how exposure reduction outside the VOI affected the image quality of the reconstructed VOI image.

The simulation was divided into six steps.

II.B.1. Noise variance versus image signal

To estimate the noise variances in the projection images, we measured the noise variance, σ^2 , and mean image signal,

S , in a 256×256 ROI at the center of the images acquired with a series of tube currents 2, 4, 6, ..., 20 mA at 80 kVp.²⁵ Noise in the projection images consists primarily of a quantum component and an electronic component. Quantum noise results from the absorption of x rays by the detector and its fluctuation follows a Poisson distribution while electronic noise originates from the image signal readout system and its fluctuation follows approximately a Gaussian distribution.²⁶ In diagnostic imaging, the photon number is usually large enough for the Poisson distribution to be approximated as a Gaussian distribution. Thus, the noise variance in image signals, σ^2 , can be expressed as:

$$\sigma^2 = \sigma_q^2 + \sigma_e^2, \quad (4)$$

where σ_q^2 and σ_e^2 are the variance of quantum and electronic noises, respectively. σ_q^2 is proportional to the number of detected photons and approximately to the mean image signal, S , while σ_e^2 characterizes the image acquisition system and is

FIG. 3. A stationary gantry, rotating phantom cone beam CT system with a VOI mask.

independent of the exposure or signal level. Therefore, we assumed the relationship between σ^2 and S was a linear function, as follows:

$$\sigma^2 = C_1 \times S + \sigma_e^2. \quad (5)$$

After iterative curve fitting, $C_1=0.04$ and $\sigma_e^2=32.56$. The function of the fitting curve can be written as²⁵

$$\sigma^2 = 0.0352S + 32.5541. \quad (6)$$

In Fig. 4, the measured data are plotted as a function of the mean image signal along with the fitted curve. The crosses and solid line represent the measured data and fitted curve, respectively. Equation (6) was used to estimate the noise variance for an image signal level.

II.B.2. Noise variances in low exposure images

To simulate a reduced exposure image, the image signals were first multiplied by a factor α , which was smaller than 1. If S and S' denote the original image signal and reduced exposure image signal, the relationship between them can be written as:

$$S' = \alpha S. \quad (7)$$

Consequently, the noise variance in the down-scaled signals was also reduced to $(\alpha\sigma)^2$. However, the noise level did not decrease as fast as the signal did because σ_e was stationary while σ_q was approximately proportional to $\sqrt{S}$. The actual noise variance can be written as:

$$\sigma'^2 = \sigma_e^2 + \alpha\sigma_q^2. \quad (8)$$

Thus, the noise variance in the down-scaled signals, $(\alpha\sigma)^2 = \alpha^2\sigma_e^2 + \alpha^2\sigma_q^2$, was smaller than that in the signals generated with an exposure reduced by the factor of α .

II.B.3. Variance of noise to be added ($\Delta\sigma^2$)

To simulate the actual VOI scanning data, an additional noise image should be generated and added to the down-scaled image data. The variance of the added noise, $\Delta\sigma^2$, may be estimated from those in the actual and down-scaled image data, as follows:

$$\Delta\sigma^2 = \sigma'^2 - (\alpha\sigma)^2. \quad (9)$$

Using Eqs. (4) and (8), $\Delta\sigma^2$ may be rewritten as:

$$\Delta\sigma^2 = \alpha(1 - \alpha)\sigma^2 + (1 - \alpha)\sigma_e^2. \quad (10)$$

II.B.4. Generation of a noise image corresponding to $\Delta\sigma^2$

To generate the additional noise image from the existing projection image, σ^2 was first estimated by using Eq. (6) and then used to estimate $\Delta\sigma^2$ by using Eq. (10) on a pixel-by-pixel basis. The additional noise fluctuation was generated by sampling directly from the following Gaussian distribution function:

$$P(x) = \frac{1}{\sqrt{2\pi} \cdot \Delta\sigma} e^{-\frac{x^2}{2\Delta\sigma^2}}, \quad (11)$$

where x is the random noise signal.

II.B.5. Filtering of the noise image

The additional noise image generated in step 4 was added to the down-scaled images to simulate the reduced exposure image. However, the formation of the original projection images induced the detector blurring. Thus, the computer generated noise image was filtered by a 2D detector blurring function before being added to the down-scaled projection image. The detector blurring function, $G_d(u, v)$, may be modeled as the convolution of a 2D Gaussian function with an aperture function:¹⁵

$$G_d(u, v) = G_s(u, v) \otimes R_p(u, v), \quad (12)$$

where $\otimes$ is a convolution operator; $G_s(u, v)$ and $R_p(u, v)$ are the point spread functions (PSFs) for the scintillation layer and signal averaging in the pixel area, respectively. They may be expressed as follows:

$$G_s(u, v) = \frac{1}{2\pi\sigma_s^2} \exp\left(-\frac{u^2 + v^2}{2\sigma_s^2}\right), \quad (13)$$

FIG. 4. Mean signal value versus noise variance. The cross and solid line represent the measured data and fitted curve, respectively.

$$R_p(u,v) = \frac{1}{a^2} \cdot \text{rect}(u) \cdot \text{rect}(v) = \begin{cases} \frac{1}{a^2} & |u| < \frac{1}{2}, \quad |v| < \frac{a}{2}, \\ 0 & \text{elsewhere} \end{cases} \quad (14)$$

where σ_s^2 is the variance of the Gaussian function and a is the pixel size. In our simulation, $\sigma_s = 0.061$ mm and $a = 194$ μm reported for an aSi/CsI detector with a scintillator thickness of 600 μm .^{27,28}

II.B.6. Addition of filtered noise image

After incorporating detector blurring, the noise images were added to the down-scaled projection image for regions outside the VOI on a pixel-by-pixel basis. Notice that noise signals sampled from Eq. (11) have a mean level of 0. Therefore, on the average, adding of these noise images should not alter the mean signal level. The simulated VOI projection images are composed of full exposure, less noisy signals inside the VOI, and simulated reduced exposure, noisier signals outside the VOI.

Alternatively, there is a simple approach to estimate the additional noise level suggested by the reviewer. The approach is repeating the measurement of step (1) with the VOI mask. Thus obtain empirical estimation of the noise variance under the mask. The difference of noise variance between the measurements with and without VOI mask represents the additional noise to be added.

II.C. Simulation of VOI images

To simulate VOI cone beam CT, a VOI mask with a 2 cm rectangular opening was assumed to be placed 50 cm away from the x-ray source. The exposure outside the VOI was assumed to be reduced by a factor of 10 compared to that inside the VOI ($\alpha = 0.1$). The VOI mask moved horizontally and vertically during the scan, allowing the collimated x-ray beam to point at and cover the VOI. Thus, the projection data within the VOI were kept unchanged while those outside the VOI were modified view by view as described in the previous section. To convert the projection images into transmission data, the reference images were processed in an identical manner. After normalizing the VOI projection images with the VOI reference images, the FDK algorithm was used to reconstruct 3D images for the VOI.²⁹

II.D. Imaging experiments with calcification and low contrast phantoms

To demonstrate the VOI scanning technique, a 2.0-mm-thick copper mask with a 2.4-cm-diameter circular opening, which attenuated the x-ray intensity by a factor of ~ 10 outside the VOI, was placed 50 cm away from the x-ray source. The center of the VOI mask opening was aligned with the rotating axis to generate a 3.6-cm-diam spherical VOI at the rotating center. A 10.5-cm-diam gelatin phantom, with a density slightly higher than that of fibroglandular tissue³⁰ and with two groups of size 280–300 μm calcifications (calcium carbonate grains) embedded at the center (in-

side the VOI) and near the edge (outside the VOI), was imaged. To study the improvement in contrast, a uniform 11-cm-diam cylindrical polycarbonate phantom, with five 0.6-mm-diam holes (filled with water) at the center and off center, was imaged.

II.E. Monte Carlo calculation

We performed Monte Carlo simulation using the Geant4 (GEometry ANd Tracking) package³¹ to estimate the dose distribution inside a breast phantom and the primary and scatter x-ray intensity distributions with and without the VOI mask. Geant4 is a set of C++ libraries which provide the functionality necessary to perform Monte Carlo simulations of photons or other particles traveling through matter. In our simulation, the breast was modeled as a 12-cm-diam, 10-cm-high cylinder of water, centered with the rotating axis of the cone beam CT system; 10^7 50 keV x rays were used to simulate cone beam CT imaging with and without the VOI mask. A 2.0-mm-thick copper mask with an opening was assumed to be used to deliver full exposure to a 2.6-cm-diam spherical VOI at the rotating center. Due to the rotating symmetry of the phantom, all projection views are identical. This allows us to use computation for a single projection view to estimate the dose from a complete (360°) scan by integrating and averaging the energy deposited over a series of concentric rings (2×2 mm² in cross section) of incrementing radii. This considerably reduced the computation time required. The detector was modeled as a perfect x-ray absorber divided into 5×5 mm² pixels. The input energy of x rays at the detector was tallied separately for primary and scattered x-rays and integrated as the primary and scatter image signals. The physics processes included the Photoelectric effect, Compton scattering, and Rayleigh scattering.

III. RESULTS

III.A. Simulation results of breast specimen

Figure 5 shows the simulated VOI cone beam CT images of a postmastectomy breast specimen side by side with the full-field cone beam CT images. It is found that the breast tissue structures within the VOI were clearly visible while those outside the VOI were obscured due to considerably higher noise level [Figs. 5(a)2 and 5(b)2]. Both the front view and side view images demonstrate that the image quality inside the VOI was comparable to that obtained with the unmasked projection images [Figs. 5(a)1 and 5(b)1]. No obvious artifacts were observed. In the difference image [Figs. 5(a)3 and 5(b)3] between the full-field and VOI cone beam CT images, only noise images were left while the breast tissue images were subtracted.

III.B. Experimental demonstration with calcification phantom

Figure 6 shows two slices of cone beam CT images of the calcifications embedded in the breast obtained with the VOI mask [Figs. 6(b) and 6(d)] and without [Figs. 6(a) and 6(c)]. While the attenuation by the VOI mask did not allow the

FIG. 5. Front (a) and side (c) views of full-field cone beam CT images of a breast specimen obtained at 80 kVp. Front (b) and side (d) views of VOI cone beam CT images of the breast specimen with a 2 cm rectangular opening.

calcifications outside the VOI to be clearly seen in the reconstructed image [Fig. 6(d)], those inside the VOI were clearly visible despite exposure reduction and noise increase outside the VOI [Fig. 6(b)]. Visibility of the calcifications within the VOI appeared to be similar to that of those at the center of the full-field image [Fig. 6(a)].

III.C. Noise level

Figure 7 shows the CT number profiles inside and outside the VOI image. Although the CT image signals outside the VOI were substantially noisier than those within it, the noise level inside the VOI was significantly lower and similar to that in the full-field cone beam CT image. Despite differences in the noise level, the signal levels inside and outside the VOI cone beam CT image appeared to be similar to those in the full-field images, too.

III.D. Potential dose saving and scatter reduction

In Fig. 8, the radiation dose distributions estimated for the breast phantom with and without the VOI mask are plotted as a function of the radius [distance from the VOI center in the horizontal direction, Fig. 8(a)] and the z level [distance from the VOI center in the vertical direction, Fig. 8(b)]. It is found that with the VOI scanning technique, the doses were substantially reduced both inside and outside the VOI. As the radius increases, the dose reduction factor outside the VOI increased significantly from 5.8 to 10.4. As the z level increased, the dose reduction factor outside the VOI increased

FIG. 6. Full-field cone beam CT image (a, c) and VOI cone beam CT image (b, d) of a 10.5-cm-diam gelatin phantom inserted with two groups of calcifications (size 280–300 μm) in the center and at the edge.

significantly from 13.2 to 24.9. The dose in the z direction was reduced more than that in the radial direction because the x-ray exposure was collimated in all projection views. The dose inside the VOI was also lowered by a factor of ~ 4.4 because of the VOI mask's the collimation effect of the VOI mask used and the resulting scatter reduction.

Figure 9 shows the primary and scattered x-ray intensities in the horizontal direction at the detector input. It is found that the primary intensity inside the VOI remained unchanged while those outside the VOI were substantially reduced due to attenuation by the VOI mask. The scatter intensities were substantially reduced both inside and outside the

FIG. 7. Comparison of CT number profiles of the VOI image and full-field cone beam CT image.

(a)

(b)

FIG. 8. Dose versus distance from the rotating axis. The vertical dashed line indicates the boundary of the VOI.

VOI. The scatter signals inside the VOI were reduced by a factor of ~ 7.3 due to the collimation effect of the VOI mask used.

III.E. Contrast-to-noise ratio (CNR) improvement

Figure 10 shows low contrast phantom images obtained with [Fig. 10(b)] and without [Fig. 10(a)] the VOI mask. The visibility of the low contrast object inside the VOI appeared to be greater when the VOI mask was used than when it was not. The CNRs inside the VOI were 0.88 [Fig. 10(a)] and 1.28 [Fig. 10(b)], respectively. Thus, the VOI technique could improve the CNR inside the VOI by $\sim 45\%$ for an 11-cm-diam breast tissue equivalent phantom.

IV. DISCUSSION

Our preliminary studies have demonstrated that the cone beam VOI scanning technique can potentially be used to image a preselected VOI with the same or better quality while reducing the dose to the breast both inside and outside the

FIG. 9. Primary and scattered x-ray intensity profiles for the full-field and VOI projection images in the horizontal direction.

VOI. Two approaches can be used to implement this technique. One is to center the breast on the rotating axis and then move the VOI mask to track the VOI. The other is to align the VOI with the iso-center, thus the position of the VOI mask can be kept fixed during the scan. With the first approach, it is unnecessary to reposition the patient. However, the need to control of the VOI mask's movement complicates the implementation of the technique and might result in misalignment artifacts. With the second approach, the pa-

(a)

(b)

FIG. 10. Full-field cone beam CT image (a) and VOI cone beam CT image (b) of a 11-cm-diam low contrast phantom. The CNRs inside the VOI are 0.88 and 1.28, respectively.

tient needs to be repositioned if the selected VOI is not located at the center of the breast. If the VOI is close to the periphery of a large breast, the size of the flat panel detector may not be large enough to cover the breast in all views. The detector therefore may need to be shifted for offset scanning and a detector offset reconstruction algorithm used for image reconstruction.^{15,32,33} Dose reduction and image quality with these two approaches might be different and need to be investigated in future studies.

Chityala *et al.*^{17,18} have reported their implement of the ROI cone beam CT technique on neuro-angiography and head phantom. We demonstrate its application on breast imaging. Although the VOI mask increased the noise level of the projection data outside the VOI, the noise level inside the VOI was comparable to that in the full-field cone beam CT images. The differences in CT numbers between the VOI cone beam CT images and full-field cone beam CT images were minimal, as shown in Fig. 7. This indicated that the substantially noisier projection data outside the VOI had little effect on the image quality of the reconstructed image inside the VOI, which agrees well with Chityala's¹⁸ simulation results with vessel phantom. In the FDK reconstruction algorithm, the projection data are first convolved with a Ramp filter kernel for each view angle, then backprojected to the 3D space. It was shown by Kak and Slaney³⁴ that the amplitude of the Ramp filter drops quickly in the spatial domain. This indicates that the high level noise outside the VOI will not significantly propagate into the VOI during the filtering process. However, the reconstructed CT numbers in the area close to the VOI boundary (about 4 voxels) may be affected.

The VOI mask attenuated the x-ray exposure in the region outside the VOI and reduces the scatter component both inside and outside the VOI, as shown in Fig. 9. This resulted in significantly lowered patient doses both inside and outside the VOI (Fig. 8) and improved CNRs inside the VOI (Fig. 10). The implication is that a VOI may be selectively scanned to achieve better image quality due to reduced scatter while substantially reducing the dose both inside and outside the VOI. This also allows higher exposure to be used to achieve better image quality images for the VOI without significantly increasing the patient dose outside the VOI. Chityala's¹⁸ results show the dose reduction outside the VOI was equal to the exposure reduction while ours show more reduction of dose than that of exposure. This is probably because Chityala did not include scatter component in their study.

The use of the VOI mask resulted in a nonuniform noise level as well as nonuniform primary and scatter signals distributions. The difference in scatter signal levels cannot be eliminated by simple signal scaling methods and generate sharp changes of signal level across the boundary of the VOI. These changes should be minimized prior to FDK reconstruction to avoid reconstruction artifacts. Methods to address these problems are being developed to minimize the differences in CT numbers inside and outside the VOI and artifacts near the boundary of the VOI.

The selection of a proper VOI will be challenging in applying the VOI scanning technique. However, due to dose saving and scatter reduction demonstrated in our study, we believe that the VOI scanning technique could play a potentially significant role in the diagnosis and staging of breast cancers though the details of implementing and applying the technique will need more investigations to optimize. Furthermore, the VOI scanning technique may have use in other applications as well. Although the simulation and experiment reported here are specifically for breast imaging, the results indicate that there could be even greater benefits with applications involving a larger object, such as pelvis, with a larger field of view and greater presence of scattered radiation.

V. CONCLUSION

In this study, we investigated the feasibility of using full exposure projection data inside the VOI in conjunction with reduced exposure, noisier projection data outside the VOI to reconstruct cone beam CT images for the VOI. The image quality inside the VOI was found to be comparable to that in the unmasked cone beam CT images with significantly reduced patient absorbed dose. With this technique, it is possible to selectively image a VOI with reduced scatter effect and lowered radiation dose both inside and outside the VOI.

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