The role of off-focus radiation in scatter correction for dedicated cone beam breast CT

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(Received 11 May 2017; revised 2 November 2017; accepted for publication 12 November 2017; published 16 December 2017)

Purpose: Dedicated cone beam breast CT (CBBCT) suffers from x-ray scatter contamination. We aim to identify the source of the significant difference between the scatter distributions estimated by two recent methods proposed by our group and to investigate its effect on CBBCT image quality.

Method: We recently proposed two novel methods of scatter correction for CBBCT, using a library based (LB) technique and a forward projection (FP) model. Despite similar enhancement on CBBCT image qualities, these two methods obtain very different scatter distributions. We hypothesize that the off-focus radiation (OFR) is the contributor and results in nontrivial signals in x-ray projections, which is ignored in the scatter estimation via the LB method. Experiments using a thin wire test tool are designed to study the effect of OFR on CBBCT spatial resolution by measuring the point spread function (PSF) and the modulation transfer function (MTF). A narrow collimator setting is used to suppress the OFR-induced signals. In addition, “PSFs” and “MTFs” are measured on clinical CBBCT images obtained by the LB and FP methods using small calcifications as point sources. The improvement of spatial resolution achieved by suppressing OFR in the wire experiment as well as in the clinical study is quantified by the improvement ratios of PSFs and spatial frequencies at different MTF values. Our hypothesis that OFR causes the imaging difference between the FP and LB methods is verified if these ratios obtained from experimental and clinical data are consistent.

Results: In the wire experiment, the results show that suppression of OFR increases the maximum signal of the PSF by about 14% and reduces the full-width-at-half-maximum (FWHM) by about 12.0%. Similar improvement on spatial resolution is achieved by the FP method compared with the LB method in the patient study. The improvement ratios of spatial frequencies at different MTF values without OFR match very well in both studies at a level of around 16%, with an average root-mean-square difference of 0.47%.

Conclusion: The results of the wire experiment and the clinical study indicate that the main difference between the LB and FP methods is whether the OFR-induced signals are included after scatter correction. Our study further shows that OFR significantly affects the image spatial resolution of CBBCT, indicating that the visualization of micro-calcifications is susceptible to OFR contamination. Our finding is therefore important in further improvement of diagnostic performance of CBBCT. © 2017 American Association of Physicists in Medicine [<https://doi.org/10.1002/mp.12686>]

Key words: cone-beam breast CT, off-focus radiation, scatter correction

1. INTRODUCTION

At this time, one vendor has received FDA approval for a dedicated breast CT (CBBCT) system for diagnostic use.¹ This system provides tomographic and three-dimensional (3D) images in any plane without physical compression as used in digital mammography or breast tomosynthesis. The tomographic and 3D imaging capability is expected to improve the diagnostic performance.² However, the image

quality of CBBCT is challenged by x-ray scatter contamination due to the large irradiation volume in each x-ray projection.^{3,4} The induced scatter artifacts on CBBCT images appear as shading/cupping artifacts and reduced image contrast. We recently proposed two effective scatter correction methods for CBBCT.^{5,6} In this article, we discover a significant difference between the estimated scatter distributions by these two methods and identify its source based on the physical modeling of x-ray imaging. Such a difference is

commonly seen in scatter studies of cone beam CT (CBCT) imaging.^{7–10}

Over the past decades, scatter correction has been an active research topic for CBCT due to the increasing clinical importance of this approach. Comprehensive reviews of scatter correction methods can be found in Refs. [11–13]. Current scatter correction methods can be classified into four main categories: scatter rejection using physical devices,^{12,14–17} analytical modeling,^{18–22} Monte Carlo (MC) simulation,^{23–28} and scatter measurement.^{7,10,29–32} Although these methods demonstrate success under certain conditions, a clinically optimal solution of scatter correction is yet to be established.^{12,13}

Toward scatter correction for CBBCT, we recently proposed two novel methods, using a library based (LB) technique⁵ and a forward projection (FP) model.⁶ Both methods are considered clinically practical as they are computationally efficient and require no changes on the current CBBCT imaging protocol. In the LB method, a scatter library is precomputed based on a simplified semi-ellipsoidal breast model with different sizes. The scatter distribution is first selected from the library according to the breast size and then preprocessed and subtracted from the raw projection for scatter correction. In the FP method, scatter-free primary projection is first simulated by forward projecting the binary-object image segmented from the uncorrected CBBCT, and the final scatter distribution is obtained by subtraction of the simulated primary projection from the raw measured projection and local filtration,²⁹ a specially designed technique which obtains a whole-field scatter distribution from sparse samples. Similar to the LB method, the acquired final scatter is finally subtracted from raw projections for effective scatter correction. As demonstrated in our publications,^{5,6} both LB and FP algorithms effectively remove scatter artifacts and improve image contrast on patient data.

Despite similar enhancement on CBBCT image qualities, we find out that the results obtained in FP method have a considerable improvement on spatial resolution compared with those of the LB method. A somewhat surprising phenomenon we observe on the LB and FP methods is that these two methods obtain very different scatter distributions. One example is shown in Fig. 1. It is worth noting that, in the existing literature on scatter estimation/correction for CBCT imaging, scatter distributions similar to Fig. 1(a) are commonly seen in MC-based methods,^{23,26,27} while measurement or analytical modeling based methods^{18,20,33} usually generate scatter distributions similar to Fig. 1(b). A line profile comparison in Fig. 1(c) shows a large discrepancy between the two scatter estimations outside the breast projection. This finding is very similar with the result presented in Chen's work,⁹ where a large discrepancy toward both edges of a cylinder phantom between the measurements and Monte-Carlo simulation suggest that a large scatter component may come from sources other than the phantom. In this work, we aim to identify the difference of physical modeling between the LB and FP methods that leads to different scatter estimates and to study its effect on the image quality of CBBCT. We find that the off-focus radiation (OFR) results in non-trivial signals in x-ray projections, which is ignored in the scatter estimation via the LB method. This finding is supported by experimental studies and its impact on the spatial resolution and contrast of CBBCT imaging is investigated using clinical data.

2. METHODS AND MATERIAL

2.A. LB vs. FP based scatter correction

We first briefly review the LB and FP algorithms for scatter correction in CBBCT, of which the major procedures are

FIG. 1. Example of the estimated scatter distributions by the LB (a) and the FP (b) methods. (c) comparison of one-dimensional profiles extracted from the dashed line shown in (a) and (b). Display window: [100 2000] detector counts. [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 2. Workflows of the LB and FP methods. The dashed squares highlight the major differences.

illustrated in Fig. 2. The main differences between the two methods are bounded in the dashed squares.

In the LB method,⁵ a critical step is to obtain a scatter map from a scatter library based on the estimated breast diameter. The scatter library is pre-computed using MC simulations. In general, accurate scatter estimation in CBCT imaging requires a scatter library with a huge size (therefore tremendous computational burden) to accommodate different factors that contribute to changes of scatter, including scan geometry and object distribution. The major contribution of our previous work⁵ is that we develop a practical scatter correction method for CBBCT with a very small library and therefore achieve high efficacy and efficiency. Using clinical studies, we show that a scatter library with only one input parameter of the breast size is sufficient for effective scatter

correction, due to the relative simplicity of breast geometry and composition.

A few approximations have to be in place in the LB method to simplify its implementation. An accurate distribution of the x-ray source is not only difficult to obtain but also shift variant on the detector.³⁴ As such, we use an ideal x-ray point source in the MC simulation for scatter library generation to save computation, and the scatter library therefore does not include OFR induced scatter signals. However, as we show in an earlier publication,³⁴ in the current CBCT systems, OFR contributes 10% to 15% of the total radiation.

The FP method adopts a very different strategy of scatter estimation.⁶ Instead of directly estimating the object scatter, a primary projection is simulated based on an ideal x-ray point spot and then subtracted from the raw projection for a first

FIG. 3. Experiment setups for PSF and MTF measurement with (left) and without (right) OFR.

FIG. 4. The Gaussian-fitted detector LSF with and without the OFR obtained from the wire experiment. [Color figure can be viewed at wileyonlinelibrary.com]

scatter estimate. As a result, the obtained scatter distribution includes all signal components other than the primary signals, which consist of both the object scatter and the OFR induced scatter. Note that, in the FP method, our final goal is to obtain a smooth estimate of scatter signals. The estimation of primary signals using forward projection is an intermediate

FIG. 5. Zoom-in views (matrix size of 200×200 with a pixel size of 0.0125 mm) of the reconstructed PSFs (a) with and (b) without OFR. (c) is the central line profile comparison for the two PSFs. [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 6. Comparison of the MTFs computed with and without OFR. [Color figure can be viewed at wileyonlinelibrary.com]

step, after which there are two more critical procedures: sparse sampling and local filtration. These procedures guarantee that only low-frequency signals appear in the final scatter estimate. Although theoretically we need to use a finite focal-spot distribution in the estimation of primary signals, we find negligible difference in the scatter estimation result compared with that using an ideal x-ray point spot.

The difference between the LB and FP methods can be better explained using Eq. (1) shown below. The measured raw projection in the unit of photon counts, I , contains three components, the primary signals, I_p , the object-induced scatter signals from the focal spot, I_s , and the additional scatter signals caused by OFR, I_{OFR} :

$$I = I_p + I_s + I_{OFR} \quad (1)$$

The FP method estimates scatter signals from $(I - I_p)$, and therefore I_{OFR} is removed after scatter correction, while the LB method estimates only I_s .

2.B. Effect of OFR on CBBCT

In this paper, we aim to investigate the effect of I_{OFR} on the CBBCT image quality and then to compare the imaging performances of the LB and FP methods. OFR causes blurring on projection images. We first design experiments to measure the system point spread function (PSF) and modulation transfer function (MTF) on CBBCT with and without OFR. Clinical CBBCT images with scatter correction by the LB and FP methods are then evaluated and compared on the spatial resolution.

2.B.1. PSF and MTF measurements with and without OFR

All the measurements are carried out on the tabletop x-ray CBCT system in our lab.³⁵ The major components of our tabletop system match those of a clinical CBBCT system,

FIG. 7. Left panel: Comparison of CBBCT images in coronal views, obtained without scatter correction (first column), with the LB correction (second column) and with the FP correction (third column). Display window: [0.25 0.35] cm⁻¹. Right panel: Zoom-in views of highlighted ROIs (50 × 50 pixels) in left panel without scatter correction, with the LB correction and with the FP correction. Display windows for each patient are adjusted to best present the calcifications.

which include a tungsten target x-ray tube (Varian RAD-94, Varian Medical Systems, Salt Lake City, UT, USA), a Eureka MC150-C collimator, a thallium-doped Cesium Iodine (CsI:

Tl) flat-panel detector (PaxScan® 4030CB, Varian Medical Systems, Salt Lake City, UT, USA), and a rotary stage for CT data acquisition. To mimic the setup of the clinical research

FIG. 8. 1-D profiles passing through the calcifications shown in Fig. 7, indicated as the dashed lines in the first column of Fig. 7. The small window in each plot shows the zoom-in view of calcification peaks bounded in the dashed rectangle. [Color figure can be viewed at wileyonlinelibrary.com]

CBBCT prototype system used for the clinical data acquisition, the x-ray tube is operated at 50 kVp, 80 mA with a nominal focal spot size of 0.4 mm. The distance between collimator and focal spot is about 22 cm and the source to detector distance is 150 cm. This setting gives a magnification of about 1.17 from the collimator plane to the source, which warrants the occlusion of most off-focus radiation with a narrow collimation. The detector has an active area of $40 \times 30 \text{ cm}^2$ running in a 2-by-2 binning mode with an effective detector pixel size of $0.388 \times 0.388 \text{ mm}^2$. The frame rate of detector is set at 30 frames/s.

During the experiment design, we find that it is challenging to separate the object-induced scatter and the OFR in a cone-beam projection. If a blocker-based method is used for separating the two, the blocker size must be large enough to remove the OFR (i.e., to avoid the penumbra effect). However, large blockers significantly change the incident x-ray field, and therefore the measured scatter using large blockers is different from that in a cone-beam projection without blockers. The most practical way of

suppressing the OFR is to use a small field of x-ray projection. However, the object-induced scatter decreases as well when the size of projection field is reduced, making it difficult to estimate the ground truth of the OFR. Therefore, to avoid the object-induced scatter I_s and study the effect of only I_{OFR} , we scan a thin wire to generate a line-spread function (LSF) on the detector and then perform a standard filtered backprojection (FBP). The axial view of the reconstructed image is considered as the PSF of CBBCT. In the experiment, we place a copper-nickel wire with a diameter of $127 \mu\text{m}$ on the top of detector and perform an air scan without an object. In one of our previous studies, we show that x-ray beam collimation effectively reduces OFR on the detector.³⁴ We therefore repeat the experiment using a wide and a narrow collimator setting to obtain PSFs and MTFs with and without OFR. As shown in Fig. 3, to remove most of I_{OFR} , we use a collimator width of 0.37 mm, which corresponds to an exposure size of 0.44 mm on the focal spot plane, to irradiate a small vertical region covering the thin wire. In each measurement, 200 frames of

projections are acquired and averaged to reduce the noise during data acquisition.

A one-dimensional LSF is first obtained from the projection on the thin wire and then converted into line integrals using a log operation. Each detector pixel value is considered as one sample of LSF at the location calculated as the distance from the detector pixel to the wire center in the direction perpendicular to the wire. Note that, to ensure the data are more finely sampled, we intentionally place the wire in the vertical direction with a small oblique angle of about 1.4 degrees, as seen in Fig. 3. The LSF is then generated via Gaussian fitting on these data samples, with a small sampling period of 0.025 mm, i.e., 1/15 of the detector pixel size.

To obtain the PSF of CBBCT, we assume that the calculated LSF results from the projection on a virtual wire located at the rotational center of the imaging system. Due to the rotational symmetry, the CBBCT image of the virtual wire, i.e., the PSF, is generated via FBP on 1,000 identical LSFs, using geometric parameters of a clinical CBBCT system. The PSF image has a size of 1024×1024 pixels with an isotropic pixel size of 0.0125 mm. On our clinical prototype CBBCT, the source to axis of rotation distance is 650 mm and the source to the detector distance is 898 mm, resulting in a magnification factor of 1.38. The virtual wire therefore has a size of 92 μm (i.e., 127 $\mu\text{m}/1.38$). Note that, we do not perform an actual CT scan on a thin wire to obtain the PSF. Such design has two practical issues for the investigations on the role of off-focal radiation. First, since OFR causes very small signal difference on the projection images, we have to use the trick of oblique wire placement to get an accurate measurement of the LSF on the projection image. The reconstructed CT image of the wire directly from the projection images cannot be used for the measurement of system MTF due to the issue of under-sampling. Secondly, if we place the thin wire around the isocenter on the rotatory stage of our benchtop system, it is difficult to avoid dangling or vibration during the scan. If we attach the wire to a phantom to stabilize its placement, the phantom causes additional object-induced scatter and we cannot assume that the signal difference between projections with narrow and wide collimation is only due to OFR. For the above reasons, we choose to use the virtual wire concept to generate the PSF and MTF of the CBBCT system.

The MTF computation from PSF is similar to the method used by Kwan et al.³⁶ A one-dimensional MTF is obtained via taking the magnitude of the Fourier Transform of the row-wise integration on the PSF image and then normalizing at the zero frequency.

2.B.2. Evaluation of clinical CBBCT images corrected using the LB and FP methods

The patient data presented in this work were acquired during a clinical research study under a protocol approved by the institutional review boards of the University of Rochester Medical Center and the University of Massachusetts Medical School. We first compare the image quality of the scatter

corrected CBBCT images by the LB and FP methods. Small calcifications on the images are selected as point objects, on which PSFs are directly measured. MTFs are then calculated in the similar way as shown in the previous section. Note that, the MTFs calculated from small calcifications are not the ground truth but rather the true MTF convolved with a kernel H, where the H considers the irregular shape and size of micro-calcifications. Therefore, a direct comparison of the MTF for LB and FP method is not possible by using the calcifications extracted from clinical images. Instead, we focus on investigating the improvement ratio of MTF between the two correction methods. By taking the ratio of the two “MTF” results from the LB and the FP methods, we cancel the effect of kernel H on the result and therefore a micro-calcification with an ideal spherical shape is not required in our analysis. As discussed earlier, we hypothesize that the LB and FP methods mainly differ in whether the OFR-induced signals are included after scatter correction. To verify this hypothesis, we compare the MTF ratios between the CBBCT images corrected by LB and FP with those obtained in the wire experiment. The two sets of MTF ratios are expected to match if the improvement of spatial resolution is indeed caused by the removal of OFR.

3. RESULTS

3.A. Measured PSF and MTF with and without OFR

Figure 4 shows the Gaussian-fitted detector LSFs obtained from the wire experiment. It is seen that OFR degrades the spatial resolution with a larger full-width-at-half-maximum (FWHM) of LSF. The resultant CBBCT PSFs are shown in Fig. 5. The central line profiles of the two PSFs

TABLE I. Summary of variations in FWHM, contrast and CNR of the calcifications included in the ROIs for the uncorrected, LB corrected and FP corrected images.

	Uncorrected	LB correction	FP correction	Increase percentage (%)
Patient a				
FWHM	2.43	2.43	2.16	-11.04
Contrast	0.15	0.20	0.23	17.73
CNR	30.87	35.43	37.92	7.03
Patient b				
FWHM	2.01	2.01	1.77	-12.09
Contrast	0.15	0.17	0.20	13.11
CNR	30.13	30.98	29.23	-5.65
Patient c				
FWHM	0.82	0.87	0.76	-12.34
Contrast	0.27	0.32	0.38	16.41
CNR	65.02	67.43	67.07	-0.53
Patient d				
FWHM	0.87	0.87	0.77	-11.68
Contrast	0.16	0.17	0.19	11.53
CNR	36.38	37.27	35.65	-4.33

FIG. 9. Comparison of the “MTFs” computed from the ROIs centered at calcification on the uncorrected, LB and FP corrected images. [Color figure can be viewed at wileyonlinelibrary.com]

[Fig. 5(c)] indicate that, in the absence of OFR, the FWHM of PSF decreases about 12% and the maximum signal of PSF increases about 14%.

The effect of OFR on image spatial resolution is also seen in the MTF comparison shown in Fig. 6. With OFR, the spatial frequencies at the MTF values of 50%, 10%, 5%, and 0.5% are 1.13, 2.03, 2.30, and 3.01 lp/mm respectively. Whereas without OFR, these spatial frequencies are 1.31, 2.34, 2.65, and 3.38 lp/mm, corresponding to an improvement of 16.1%, 15.5%, 15.1% and 12.1% at 50%, 10%, 5%, and 0.5% MTF values respectively.

3.B. Evaluations of clinical CBBCT images

Figure 7 compares the scatter correction results using the LB and FP methods of four patients. It is observed that although both methods effectively remove the shading artifact caused by scatter and enhance the overall image contrast (left panel of Fig. 7), the images corrected by FP appear to be

crispier. This fact is better illustrated by the zoom-in views of regions of interest (ROIs) around the calcifications (right panel of Fig. 7) and comparisons of the one-dimensional profiles (see Fig. 8) extracted from the dashed line passing through the calcifications. Table I quantitatively summarizes the variations in FWHM, contrast and CNR of these calcifications. Compared with LB, the FP method increases contrast between the maximum signals and background of the line profiles by 14.5% and reduces the FWHMs by 12% on average. And there is no significant difference on the image CNR for these two methods. If we consider the line profiles passing through calcifications as analogs to image PSFs, the above result regarding the contrast and FWHM are consistent with what we found in the wire experiment.

We further compute “MTFs” from the zoom-in ROIs shown in Fig. 7 as image “PSFs”. In Fig. 9, it is seen that the FP method achieves better image spatial resolution than the LB method overall. We also find that there is little difference between the “MTFs” for uncorrected and LB corrected

FIG. 10. Comparison of the improvement ratios of spatial frequencies at different MTF values for each patient case shown in Fig. 9 with the result of wire experiment shown in Fig. 6. [Color figure can be viewed at wileyonlinelibrary.com]

images. The result is conceivable since the scatter component in LB estimation is only low frequency therefore has limited ability in improving the spatial resolution of high contrast objects. This finding is also consistent with the result published in Ref. [37]. Note that, these “MTFs” have different shapes because the calcifications have different physical sizes. We then calculate the improvement ratio of spatial frequencies at different MTF values and compare with that obtained from Fig. 6 in the wire experiment. Figure 10 reveals that the improvement ratios are highly consistent with those of the wire experiment for all the patient cases, with an average root-mean-square difference of 0.47%.

The comparison of PSFs and MTFs show that the FP method outperforms the LB method on the image spatial resolution, and the improvement matches very well with the result of the wire experiment when OFR is suppressed. We therefore conclude that OFR causes the imaging difference between the FP and LB methods.

4. CONCLUSIONS AND DISCUSSION

In this work, we compare two previously proposed scatter correction methods (i.e., the LB and FP methods) and observe a significant difference in the scatter estimation. We hypothesize that the large discrepancy in the obtained scatter distributions is due to OFR, which is effectively removed only by the FP method, leading to improved image spatial resolution. To verify our hypothesis, a wire experiment is first designed to quantify the effect of OFR on the image spatial resolution. The results show that suppression of OFR increases the maximum signal of PSF by about 14% and reduces FWHM by about 12.0%. Similar improvement on spatial resolution is achieved by the FP method compared with the LB method in the study of four patients. The ratios of spatial frequencies at different MTF values with and without OFR match very well in both studies at a level of around 16%, with an average root-mean-square difference of 0.47%.

This result serves as a strong evidence that the FP method outperforms the LB method on image spatial resolution, mainly by removing OFR-induced signals. Another merit of the FP method reflects as it potentially removes beam hardening artifact even in the presence of OFR. Since an ideal focal spot without OFR and with a mono-energetic spectrum is used in the forward projection process, the estimated primary signals do not contain errors from beam-hardening and OFR. The difference between the raw projection signals and the estimated primary signals is considered to include all the errors, which are mostly removed after scatter correction. It is worth mentioning that the removal of beam hardening errors changes the performance of the spatial resolution of the FP method.³⁸ However, the presented results show that the improvement ratio of MTF between the LB and FP methods matches well with that of the wire experiments with and without OFR. As such, we argue that the influence of beam hardening effect on MTF is much smaller than that of OFR in the FP method.

Although the presented work is focused on comparisons of two recent scatter correction methods, our finding reveals an important fact that OFR, an error source considered as secondary in x-ray imaging, results in substantial difference of imaging performance in CBBCT. A critical requirement for a breast cancer diagnostic device is the ability to accurately detect micro-calcification as they are frequently associated with malignancy.^{39,40} To date, many efforts have been exerted to optimize the detector performance of CBBCT system for better detection of micro-calcifications,^{41,42} but there have not been any studies to investigate the dependency of OFR. The results presented in this study show that the OFR has a significant impact on the image quality in terms of spatial resolution and contrast, and this effect is especially prominent for fine structures (i.e., micro-calcifications) in CBBCT images. We believe that our finding is potentially important for enhancing the diagnostic performance of CBBCT in terms of calcification detection.

ACKNOWLEDGMENTS

Research reported in this publication was partially supported by the National Institute of Biomedical Imaging and Bioengineering, and the National Cancer Institute, of the National Institutes of Health under Award Numbers R21EB019597, R21EB021545, R21CA134128, CA199044, and R01 CA195512. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health. The project was also partially supported by the Ministry of Science and Technology of China Key Research and Development Project (Grant No. 2016YFC0101400), the National Natural Science Foundation of China (Grant No. 81671681) and the Fundamental Research Funds for the Central Universities.

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