A software-based x-ray scatter correction method for breast tomosynthesis

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Purpose: To develop a software-based scatter correction method for digital breast tomosynthesis (DBT) imaging and investigate its impact on the image quality of tomosynthesis reconstructions of both phantoms and patients.

Methods: A Monte Carlo (MC) simulation of x-ray scatter, with geometry matching that of the cranio-caudal (CC) view of a DBT clinical prototype, was developed using the Geant4 toolkit and used to generate maps of the scatter-to-primary ratio (SPR) of a number of homogeneous standard-shaped breasts of varying sizes. Dimension-matched SPR maps were then deformed and registered to DBT acquisition projections, allowing for the estimation of the primary x-ray signal acquired by the imaging system. Noise filtering of the estimated projections was then performed to reduce the impact of the quantum noise of the x-ray scatter. Three dimensional (3D) reconstruction was then performed using the maximum likelihood-expectation maximization (MLEM) method. This process was tested on acquisitions of a heterogeneous 50/50 adipose/glandular tomosynthesis phantom with embedded masses, fibers, and microcalcifications and on acquisitions of patients. The image quality of the reconstructions of the scatter-corrected and uncorrected projections was analyzed by studying the signal-difference-to-noise ratio (SDNR), the integral of the signal in each mass lesion (integrated mass signal, IMS), and the modulation transfer function (MTF).

Results: The reconstructions of the scatter-corrected projections demonstrated superior image quality. The SDNR of masses embedded in a 5 cm thick tomosynthesis phantom improved 60%–66%, while the SDNR of the smallest mass in an 8 cm thick phantom improved by 59% ($p < 0.01$). The IMS of the masses in the 5 cm thick phantom also improved by 15%–29%, while the IMS of the masses in the 8 cm thick phantom improved by 26%–62% ($p < 0.01$). Some embedded microcalcifications in the tomosynthesis phantoms were visible only in the scatter-corrected reconstructions. The visibility of the findings in two patient images was also improved by the application of the scatter correction algorithm. The MTF of the images did not change after application of the scatter correction algorithm, indicating that spatial resolution was not adversely affected.

Conclusions: Our software-based scatter correction algorithm exhibits great potential in improving the image quality of DBT acquisitions of both phantoms and patients. The proposed algorithm does not require a time-consuming MC simulation for each specific case to be corrected, making it applicable in the clinical realm. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3659703]

Key words: digital breast tomosynthesis, x-ray scatter, Monte Carlo simulation, image quality, breast cancer, mammography

I. INTRODUCTION

Digital breast tomosynthesis (DBT) imaging¹ is emerging as a potentially feasible replacement or adjunct to standard mammography for the screening and diagnosis of breast cancer. This imaging method allows for the generation of pseudo-three-dimensional (3D) images of the breast, which can provide greater detail about the structure of the tissue, and overcomes one of the main limitations of mammography, which is the compression of information of a 3D structure into a two-dimensional image. In DBT, multiple low dose projections of the breast are acquired over a limited

angular range; these projections are then used to calculate a pseudo-3D reconstruction of the imaged tissue.^{1–3} Preliminary studies have shown that DBT has the potential to improve on the clinical performance of breast cancer screening and/or diagnosis.^{4–7}

Current tomosynthesis systems, including the clinical prototype Selenia Dimensions (Hologic, Inc., Bedford, MA) used in this study, lack x-ray scatter reduction measures, be it in software or hardware, leading to the inclusion of the entirety of the x-ray scatter signal in the tomosynthesis projections. In mammography, an antiscatter grid oriented toward

the fixed x-ray source lowers the ratio of the number of scattered x-rays incident on the detector to that of the number of incident primary (nonscattered) x-rays, or the scatter-to-primary ratio (SPR). In DBT, multiple projections are acquired as the x-ray source rotates around the breast, which prevents the use of an antiscatter grid due to the severe cut-off of primary x-rays by the grid in the non-normal projections. Therefore, as we have previously shown, in DBT the SPR can be as high as 1.6.⁸ It has also been shown that if the scatter x-rays are not removed or otherwise accounted for before tomosynthesis reconstruction, the resulting volume exhibits cupping artifacts, reduced accuracy in reconstruction values, and reduced contrast.^{9–11}

Here, we propose a software-based x-ray scatter correction algorithm that is applied to the acquired tomosynthesis projections before reconstruction. The strength of the proposed algorithm lies in a precomputed “library” of SPR maps for homogeneous standard-shaped breasts of various sizes that can be used to scatter-correct any acquired patient DBT image without performing a case-specific and time-intensive Monte Carlo (MC) simulation. We have applied the proposed method to tomosynthesis acquisitions of both phantoms and patient data and studied the resulting image reconstructions. By comparing the image quality of the original uncorrected reconstructions to that of the scatter corrected reconstructions, we have determined the effectiveness of our proposed method and its ability to potentially improve the clinical performance of DBT when its use is widespread.

II. MATERIALS AND METHODS

The proposed scatter correction algorithm is comprised of four distinct steps: retrieval of a previously computed MC simulation of a SPR map of a geometrically similar DBT acquisition, registration of the SPR map to the acquired breast projections, removal of the low-frequency scatter signal, and noise filtering of the projections. The retrieved SPR images are computed from a MC simulation implemented in C++ based on the Geant4 Monte Carlo simulation toolkit,^{12,13} similar to that used in Sechopoulos *et al.*^{3,8,14} The simulations generate two images, comprised of the incident scattered and primary x-ray energy, which are combined to form a scatter-to-primary (SPR) map. This SPR map is then deformed and registered to the acquired breast projections. The registered SPR map and the original projection are used to estimate the primary signal, henceforth known as a scatter corrected projection. Lastly, the scatter corrected projections undergo noise filtration and 3D reconstruction.

This procedure was tested on multiple tomosynthesis projection sets acquired by a clinical prototype breast digital tomosynthesis system (Selenia Dimensions, Hologic, Inc., Bedford, MA) installed at Emory University. The first tests were performed on images of a heterogeneous background Model 020 BR3D Tomosynthesis phantom (CIRS, Inc., Norfolk, VA), including a “target” plate with representations of embedded spheroid masses, fibers, and microcalcification clusters. Subsequently, as a preliminary study, the scatter

correction algorithm was performed on acquisitions from two distinct patients, one in which a mass is present and one in which microcalcifications are present. Given the limited number of cases included in this study, only SPR maps with geometry matching these cases were generated from the MC simulation. In the future, a comprehensive library of SPR maps of standard shaped breasts can be assembled, and the corresponding SPR maps would be selected. After correction, all cases were reconstructed using the maximum likelihood-expectation maximization (MLEM) reconstruction algorithm,² although any reconstruction algorithm could be used.

II.A. Monte Carlo simulations

We developed our C++ implementation of MC simulations to match the acquisition geometry of the Selenia Dimensions system for the cranio-caudal (CC) view. The system acquires 15 projections over a 15° angular range, with the x-ray tube positioned directly perpendicular to the image detector for the middle (tomosynthesis angle = 0°) projection. The simulation includes the x-ray source, the image detector, the breast support plate, and the breast compression paddle. The x-ray tube is modeled as a point source with the center of rotation located at the surface of the detector. Complete details of the system have been previously published in Ren *et al.*¹⁵ The source to imager distance (SID) for the center projection is set at 70 cm (Fig. 1). The detector measures 24 × 28 cm, with a pixel pitch of 70 μm. For each test case, the following parameters are determined from the system and the center projection: x-ray tube voltage (automatically selected by the system), compressed breast thickness (Th) (displayed by the system), chest-to-nipple distance (CND), and length along the chest wall (LCW) (both from the center projection). The compressed breast is modeled as a truncated semi-ellipsoid of a homogeneous 50%/50% mixture of adipose and glandular tissue positioned at the chest wall edge of the detector with dimensions similar to the case to be corrected (Fig. 2). When simulating breast phantoms as opposed to patients’ breasts, the breast model is altered such that it has constant thickness at the outer edge, away from the chest wall, to match the straight edges of the phantoms.

For each test case, we performed simulations of five hundred million (5.0×10^8) monoenergetic x-rays emitted from the x-ray tube toward the detector for each energy level from 10.0 keV to the maximum energy level of the spectrum selected by the tomosynthesis prototype, in 0.5 keV steps. Each photon that arrives at the detector face was recorded as a primary x-ray, or as a scatter x-ray if it experienced at least one Rayleigh and/or Compton scattering event. Since it is known that x-ray scatter varies slowly over the detector, a pixel pitch of 7.0 mm was used in the MC simulations to decrease the simulation time. The simulated scatter monoenergetic images were combined into estimated scatter spectral images using x-ray spectrum models calculated according to the method described by Boone *et al.*¹⁶ and noise was reduced using the method described by Colijn and

FIG. 1. The breast tomosynthesis system acquires 15 projections (of which only 3 are shown here), spaced evenly, from -7.5° to $+7.5^\circ$ of the central line perpendicular to the image detector plane, located at the edge of the detector next to the chest wall.

Beekman.¹⁷ To obtain noise-less x-ray primary images, ray-tracing¹⁸ was performed to determine the pathlength of each ray from the source to each pixel on the image detector. The pathlengths of the rays through each material in the

simulation combined with the attenuation coefficients for each material¹⁹ and x-ray energy were used to calculate the estimated primary spectral images. This allows for the calculation of a scatter-to-primary (SPR) map, which depicts the

FIG. 2. Monte Carlo simulations of the breast in the CC view and system geometry include the image detector, measuring 24×28 cm, with 7.0 mm pixels, the support plate, the compressed breast, and the compression paddle. The breast parameters of CND, LCW, and compressed breast thickness (Th) are determined for each case and input as parameters to the simulation. The compressed breast depicted here has curved edges opposite the chest wall, resembling patient breasts. For the simulations for the tomosynthesis phantom, the simulated breast model had straight edges.

ratio of the scatter signal to the primary signal on a pixel by pixel basis, for each tomosynthesis acquisition projection. As a final step, the SPR map is resampled to the pixel pitch of the imaging system (140 μm).

II.B. Deformation and registration of scatter-to-primary map

In the next step of our proposed scatter correction algorithm, the SPR maps obtained from MC simulations are deformed and registered to the acquired projections to match the breast tissue edge. This is crucial as the MC simulation utilizes a standardized breast shape, so its results have to be matched to the shape of the actual breast images that are being corrected. The small variability of the x-ray scatter signal with glandular fraction and breast size⁸ allows for the use of a SPR map of a homogeneous standard-shaped breast of similar, but not equal, size to the one being corrected, after registration to the edges of the acquired image. In this way, a time-consuming MC simulation does not have to be performed for each DBT case acquired.

To perform the registration, the simulated scatter and primary x-ray images are combined to generate a total estimate (T_e) of the acquired projection (T_{acq}). The edge of the breast and the chest wall in T_{acq} is located by thresholding, using Roberts cross gradient operators, and edge thinning.^{20–22} Points along the edge and chest wall are chosen, defining an outer border. For each of these outer points, an inner point is defined 1 cm inward, parallel to the image gradient, of the breast tissue projection. The outer and inner borders of the breast and the center of mass are used in a thin-plate spline interpolation registration of T_e to T_{acq} in order to determine the transformation (R) necessary to match the shape and position of the of these two breast images.²³ This transformation, consisting of a deformation and rotation and translation, is then applied to the SPR map to generate a registered SPR map ($R\{SPR\}$).

For any signal T that consists of the sum of a scatter signal (S) and a primary signal (P), it is known that S , SPR, and T are related according to the equation (see Appendix)

$$S = T \left(\frac{SPR}{1 + SPR} \right). \quad (1)$$

In this application, as the scatter signal in tomosynthesis has been shown previously to be composed primarily of low frequency signal, T_{acq} needs to be low-pass filtered before it is used in Eq. (1). Therefore, a low-pass filtered version of T_{acq} (T_{acq-LP}) is used to calculate the registered estimated scatter images (S_e) using the following equation:

$$S_e = T_{acq-LP} \left(\frac{R\{SPR\}}{1 + R\{SPR\}} \right). \quad (2)$$

The final registered primary x-ray estimates (P_e) are then calculated by subtracting S_e from T_{acq} .

II.C. Noise filtration and image reconstruction

Before reconstructing the breast tomosynthesis image, the P_e images undergo noise-filtering to reduce the impact of the

quantum noise of the scattered x-rays, since the previous steps only correct for the low-frequency offset from the scatter signal. For this, a 3×3 pixel kernel adaptive means filter is utilized, with a minimum variance level 0.25 times the noise variance level measured from tomosynthesis image acquisitions of matched thickness phantoms representative of homogenous 50%/50% glandular/adipose tissue (CIRS Model 082).²⁴

Finally, the DBT reconstruction is performed using the noise filtered P_e . For this study, the MLEM iterative method described by Wu *et al.* was used, as this method has been shown to be very effective at balancing the image quality of both masses and microcalcifications.^{2,25} However, the corrected projections resulting from our proposed method can be used for reconstruction with any algorithm.

II.D. Image quality comparison

In order to determine the benefits realized by the proposed scatter correction algorithm, the complete process described above was performed on acquisitions of a phantom and patient cases, and the image quality of the resulting scatter-corrected reconstructed 3D images was compared to that of the original, uncorrected reconstructions using image contrast metrics such as maximum signal difference (SD), signal-difference-to-noise ratio (SDNR), and integrated mass signal (IMS). In addition, to better understand the impact that each portion of the scatter correction algorithm has on the reconstructed image quality, reconstructions of the phantom projections after undergoing only the x-ray scatter quantum noise filtration (T_F) were also obtained and evaluated. The datasets are all reconstructed to image stacks of 0.14 mm $\times$ 0.14 mm $\times$ 1.0 mm voxels.

The patient case datasets were acquired for an unrelated IRB-approved clinical study from which the images were released for use in other research projects. For the phantom cases, the CIRS Model 020 BR3D Tomosynthesis phantom was utilized, consisting of 1 cm thick slabs of a heterogeneous 50%/50% mixture of adipose and glandular tissues. This phantom represents real tissue better than a homogenous phantom and tests tomosynthesis' ability to suppress tissue superposition. One of the 1 cm thick slabs, the "target slab," has embedded spheroid tumor masses, fibers, and microcalcification clusters of differing sizes (Fig. 3). Phantoms measuring 5 cm and 8 cm thick were imaged, with the target plate located at heights of 3 cm and 5 cm, respectively, and the image quality of the 3D images was evaluated using three metrics.

The maximum SD and SDNR of the spheroid mass targets were calculated by measuring the maximum and minimum signal of a region of interest (ROI) centered on each mass and the noise of a designated area of background signal in the focal plane. The circular ROI around the spheroid mass was chosen such that its diameter spanned 60 voxels and encompassed each mass entirely. Therefore, the SD and SDNR were calculated as follows:

$$SD = I_{max} - I_{min}, \quad (3)$$

FIG. 3. (a) The CIRS Model 082 breast mammography phantom, representing a 50/50 heterogeneous mixture of glandular and adipose tissue contains three types of embedded targets of varying sizes: (b) microcalcification clusters, (c) fibers, and (d) spheroid masses.

$$SDNR = \frac{I_{\max} - I_{\min}}{\sigma_{BG}}, \quad (4)$$

where $I_{\max}$ is the maximum voxel value in the ROI, $I_{\min}$ is the minimum voxel value in the ROI, and σ_{BG} is the standard deviation of voxel values in the background area, which is a fixed circular region with a diameter of 25 voxels in the focal plane, chosen for its homogeneity.

Another quantitative measure of the image quality of the spheroid mass targets is the gray level signal profile measured in the direction perpendicular to the chest wall. From this profile, the IMS was calculated for each mass. We define IMS as the area under the signal profile curve above the signal at the edge of the mass itself, where the minimum gray levels are found (Fig. 4). The IMS therefore represents the combined signal difference of the mass itself from its immediate surroundings.

To determine if the scatter correction algorithm has a significant impact on the image contrast of the spheroid masses, the Wilcoxon signed-rank test was used. The test was applied to the SD, SDNR, and IMS for all masses in both the 5 cm and 8 cm thick phantoms, to compare differences in each measure between the original reconstructions and either test case: (i) the reconstructions of the scatter corrected projections or (ii) the reconstructions of noise filtered-only projections. In both cases, the null hypothesis is that the median difference of the image contrast measurement, across the masses, between the original reconstruction and the test case

FIG. 4. (a) To measure the IMS between the mass edges, gray level signal profiles of each of the spheroid masses were taken in the direction perpendicular to the chest wall. (b) The gray level signal is offset from the minimum signal level of the profile, which is found at the edge of the spheroid mass. The IMS is the area (grey) under the offset profile curve.

is zero. The Wilcoxon signed-rank test was chosen over a paired t-test as the distribution of differences could not be assumed to be normally distributed.

To ensure that the resolution of the DBT reconstructions was not compromised by the noise filtration process, the modulation transfer function (MTF) of the projections before and after undergoing filtering was measured. The line response function (LRF) of the projections of the system was measured using a TX-5 tungsten edge (Scanditronix Wellhöfer, Schwarzenbruck, Germany), following the method of Kyprianou *et al.*²⁶ The MTF was calculated by taking the Fourier transform of the LRF.²⁷ These projections then underwent noise filtration with the filter's variance parameter set to 1, 0.5, and 0.25 times the noise variance level, the lowest level ($0.25\times$) being the actual level used in the noise filtration process. The MTF of the noise filtered projections was compared to that of the unfiltered projections.

III. RESULTS

III.A. Scatter correction of phantom images

Reconstructed slices of the focal plane of the target plate of the 5 cm and 8 cm thick phantoms are shown in Figs. 5 and 6, respectively. Figure 5(a) depicts the reconstruction of the original uncorrected projections (*O*), while Fig. 5(b) shows a zoomed-in view of the second largest of the spheroid masses (M2) and Fig. 5(c) shows a zoomed-in view of the fourth largest microcalcification cluster (C4). Figures 5(d)–5(f) depict the same ROIs, respectively, of the reconstruction of the noise filtered T_F projection set (*F*), while Figs. 5(g)–5(i) depict the same ROIs, respectively, of the reconstruction of the fully scatter corrected projection set (*C*) of the 5 cm thick phantom. All images are displayed with equal window width but different window levels. In Fig. 5(i), white arrows point to two microcalcifications of the C4 set which can be clearly seen in the *C* reconstruction but not in the *O* reconstruction [Fig. 5(c)] and to a lesser extent in the *F* reconstruction [Fig. 5(i)]. Figure 6 depicts similar views of the three reconstructions of the 8 cm thick phantom, with the zoomed-in views centered on the fourth largest spheroid mass (M4) and C4. In Fig. 6(i), white arrows point to two microcalcifications of the C4 set which, like in the 5 cm phantom, can be clearly seen in the *C*

FIG. 5. (a) Uncorrected in-plane slice of the focal plane of the embedded target plate located at the center of the 5 cm thick heterogeneous phantom, with marked regions of interest (ROI). (b) Uncorrected zoomed-in view of the ROI centered on the second largest of the spheroid masses (M2). (c) Uncorrected zoomed-in view of the ROI centered on the fourth largest microcalcification cluster (C4). White arrows point to microcalcifications that are present but not visible. (d) Noise-filtered slice of the embedded target plate of the 5 cm thick phantom, with boxed ROIs. (e) Noise-filtered zoomed-in view of the ROI centered on M2. (f) Noise-filtered zoomed-in view of the ROI centered on C4. (g) Scatter corrected slice of the embedded target plate of the 5 cm thick phantom, with boxed ROIs. (h) Scatter corrected zoomed-in view of the ROI centered on M2. The SDNR of this mass exhibited a 65% increase following scatter correction. (i) Scatter corrected zoomed-in view of the ROI centered on C4. White arrows point to microcalcifications not visible in (c) that are visible here. These images are displayed with equal window width but different window levels.

reconstruction but not in the O reconstruction [Fig. 6(c)] and to a lesser extent in the F reconstruction [Fig. 6(f)].

Quantitative analysis of the reconstructions of the O , F , and C sets revealed that the proposed scatter correction algorithm improves the SD (Table I), the SDNR (Table II), and the IMS (Table III) of the spheroid masses located in the target plate. The C reconstructions of the 5 cm thick phantom exhibited a 60%–66% improvement in SD and SDNR across the four largest spheroid masses (M1–M4). The C reconstructions of the 8 cm thick phantom exhibited a more modest improvement (17%–32%) in SD and SDNR. The improvements in both metrics were statistically significant (SD: Wilcoxon's $W=0$, $p<0.01$, $n=8$; SDNR: $W=0$,

$p<0.01$). Note that the C/O ratio of SD is very similar to the C/O ratio of SDNR because the measured noise of the C reconstructions approaches that of the O reconstructions. Although in most cases noise filtering only improves image quality, the fully scatter-corrected reconstructions exhibit the greatest improvement in both SD and SDNR. The C reconstructions also demonstrate improvements of 15%–62% in IMS ($W=0$, $p<0.01$), whereas the F reconstructions exhibit decreased IMS measurements ($W=0$, $p<0.01$). This can also be seen in the offset signal profiles displayed Fig. 7, from which the IMS is calculated. Note also that the C/O ratios of IMS measurements for all four spheroid masses of the 8 cm thick phantoms are greater than the corresponding

FIG. 6. (a) Uncorrected in-plane slice of target plate at center of 8 cm thick phantom, with boxed ROIs. (b) Uncorrected zoomed-in view of the ROI centered on the fourth largest spheroid mass (M4). (c) Uncorrected zoomed-in view of the ROI centered on C4. White arrows point to microcalcifications that are present but not visible. (d) Noise-filtered slice of the embedded target plate of the 8 cm thick phantom, with boxed ROIs. (e) Noise-filtered zoomed-in view of the ROI centered on M4. (f) Noise-filtered zoomed-in view of the ROI centered on C4. (g) Scatter corrected slice of the embedded target plate of the 8 cm thick phantom, with boxed ROIs. (h) Scatter corrected zoomed-in view of the ROI centered on M4. The SDNR of this mass exhibited a 26% increase following scatter correction. (i) Scatter corrected zoomed-in view of the ROI centered on C4. White arrows point to microcalcifications not visible in (c) that are visible here. These images are displayed with equal window width but different window levels.

C/O ratio of IMS for the 5 cm thick phantom. The two smallest masses (M5 and M6) cannot be clearly seen in any of the reconstructions and therefore were excluded from the analysis.

The MTF of the central projection of a DBT acquisition was computed and can be seen in Fig. 8. The MTF of the

TABLE I. Ratio of maximum SD of spheroid masses in heterogeneous tomosynthesis phantom between the original reconstructions (O) and the noise-filtered-only (F) and scatter-corrected (C) reconstructions. Wilcoxon signed-rank test applied across masses M1–M4 in both phantoms revealed that the differences seen in SD between the C reconstructions and the O reconstructions was significant (Wilcoxon's $W = 0$, $p < 0.01$, $n = 8$), while the differences in SD between the F reconstructions and the O reconstructions were not significant ($W = 14$, $p > 0.2$).

Phantom thickness (cm)	Reconstruction set	M1	M2	M3	M4
5	F/O Ratio	1.10	1.15	1.09	1.19
5	C/O Ratio	1.60	1.65	1.66	1.65
8	F/O Ratio	0.75	0.83	0.82	0.78
8	C/O Ratio	1.17	1.31	1.32	1.26

TABLE II. Ratio of SDNR of spheroid masses in heterogeneous tomosynthesis phantom between the original reconstructions (O) and the noise-filtered-only (F) and scatter-corrected (C) reconstructions. The differences seen in SDNR between the C reconstructions and the O reconstructions and between the F and O reconstructions were both significant (Wilcoxon's $W = 0$, $p < 0.01$).

Phantom thickness (cm)	Reconstruction set	M1	M2	M3	M4
5	F/O Ratio	1.46	1.53	1.46	1.59
5	C/O Ratio	1.60	1.65	1.66	1.65
8	F/O Ratio	1.13	1.24	1.23	1.17
8	C/O Ratio	1.17	1.31	1.32	1.26

system's projections did not differ before and after four degrees of noise filtration was applied to the projections (Fig. 8). Thus, the noise filtration did not alter the resolution of the Selenia Dimensions system.

III.B. Patient cases

In-plane views of the O and C reconstructions of the first patient case (patient A), centered around a mass anterior to the nipple, are displayed in Figs. 9(a) and 9(b), respectively. In-plane views of the O and C reconstructions of the second patient case (patient B), centered on microcalcifications at two different depths, are displayed in Fig. 10. Figures 10(a) and 10(b) depict the O and C sets, respectively, of three microcalcifications located near the x-ray tube side of the breast, while Figs. 10(c) and 10(d) depict the same of a single larger microcalcification located near the detector side of the breast. These images are also displayed with equal window width but different window levels.

IV. DISCUSSION

In this study, we proposed a Monte Carlo simulation-based scatter correction algorithm for breast tomosynthesis and compared the image quality of both phantom and patient reconstructions before and after the acquired projections were scatter corrected. Contrast enhancement can be seen in the image reconstructions of the phantoms of differing thicknesses shown in Figs. 5 and 6, where the swirl patterns of the heterogeneous mixture of 50/50 glandular/adipose tissue are more distinguished in the scatter corrected images. The zoomed-in views of the embedded spheroid masses in Figs. 5 and 6 and the improvements in SDNR shown in Table II

TABLE III. Ratio of IMS of spheroid masses in heterogeneous tomosynthesis phantom between the original reconstructions (O) and the noise-filtered-only (F) and scatter-corrected (C) reconstructions. The differences seen in IMS between the C reconstructions and the O reconstructions and between the F and O reconstructions were both significant (Wilcoxon's $W = 0$, $p < 0.01$).

Phantom thickness (cm)	Reconstruction set	M1	M2	M3	M4
5	F/O Ratio	0.85	0.75	0.88	0.91
5	C/O Ratio	1.19	1.15	1.22	1.29
8	F/O Ratio	0.79	0.90	0.93	0.83
8	C/O Ratio	1.26	1.28	1.62	1.30

FIG. 7. Offset gray level signal profiles of spheroid masses measured from original (*O*), filtered (*F*), and scatter corrected (*C*) reconstructions—the signal profiles, measured in the direction perpendicular to the chest wall, of the masses M1–M4 of the target plate of the 5 cm thick phantom can be seen in (a), (b), (c), and (d), respectively, while those of the 8 cm thick phantom can be seen in (e), (f), (g), and (h). The area under the profile curve, or IMS, of the *C* reconstructions (grey) is greater than that of the *O* reconstructions (black), which in turn is greater than that of the *F* reconstructions (dashed black). This suggests that the masses are easiest to distinguish from their surroundings in the *C* reconstructions.

and the improvements in IMS shown in Table III, both statistically significant, also support this conclusion. While the scatter corrected reconstructions of the 8 cm thick phantom do not exhibit greater improvements in SDNR than those of the 5 cm thick phantom, they do exhibit greater improvements in IMS. As DBT is not a quantitative imaging technique, we do not believe that the SDNR values of the reconstructed images can be reliably compared between different-sized images acquired with differing x-ray spectra. Instead we propose that the IMS, calculated from the offset gray level signal profiles seen in Fig. 7, allows for a better comparison of the visibility of the spheroid masses between different reconstructions of the same set of projections. In addition, the zoomed-in views of the C4 microcalcification group in Figs. 5 and 6 (white arrows) show that at least two of the microcalcifications can be clearly seen in the scatter corrected reconstructions but not in the original uncorrected set. The fibers also exhibit improved contrast after scatter cor-

rection, as they are easier to detect in both phantoms. Improvement in the visibility and detail of the lesions in the patient images were also seen, with no introduction of artifacts.

To determine if the noise filtering used to reduce the impact of the scatter x-rays quantum noise negatively impacts image resolution, we computed and compared the MTF of the central projection. The MTF was not reduced after the central projection underwent various degrees of noise filtration (Fig. 7). Thus, we believe that there is no change in the spatial resolution of the images following the application of our scatter correction algorithm.

Examination of the patient images also reveals improvements in image quality. The mass in patient A image is brighter and more easily identifiable in the scatter corrected reconstructions and greater detail is also visible inside the mass itself (Fig. 9). After scatter correction, the microcalcifications in patient B images shown in Fig. 10 are also brighter and easier to visualize. In particular, the topmost lesion in

FIG. 8. Comparison of MTF before and after noise filtration—the MTF of the projections in the direction parallel to the x-ray tube movement before and after undergoing differing levels of noise filtration. The filter's variance parameter was set to 0.25, 0.5, and 1.0 times the noise variance level.

Fig. 10(b) underwent the greatest improvement, which is very encouraging as it is also the smallest lesion.

The improvements in image quality realized by the proposed scatter correction algorithm cannot only aid clinicians in breast cancer diagnosis and screening, but it could open avenues for new breast tomosynthesis acquisition techniques. Recent studies have suggested that reduced compression of breast tissue to alleviate pain during tomosynthesis imaging might be possible without compromising image quality and without increasing patient dose.^{28,29} We have performed preliminary studies to evaluate the feasibility of reduced breast compression in breast tomosynthesis with adjustment of acquisition parameters to maintain or lower tissue dose and have found that image quality might be affected by the increased x-ray scatter signal due to the increased tissue thickness and the change in x-ray spectrum. Subsequent application of the proposed scatter correction algorithm showed great promise in restoring or even enhancing the image quality of the reduced compression acquisitions.

Lastly, the proposed method has been tested extensively on CC views of multiple phantoms and of multiple patient cases acquired by the clinical prototype installed at Emory

University. We anticipate no problems in adapting the Monte Carlo simulation geometry to match other tomosynthesis systems or a finalized clinical system. However, we would like to consider improving the representation of the simulated breast tissue. Here, we have used the cut out center of a semi-ellipsoid, generating a symmetrically rounded edge of the breast tissue. In reality, compression of breast tissue probably does not confer such a perfect symmetry to the tissue edge and could affect the scatter estimates in that region. In addition, we will continue to develop the method to accommodate the mediolateral oblique (MLO) view of DBT systems. While the MC simulation can generate SPR maps of projections acquired in the MLO view, enhancement of the deformation and registration process is required in order to account for the pectoralis muscle included in these projections. We have also shown the potential benefits of the proposed scatter correction method on two distinct clinical acquisitions to demonstrate the possible improvements in image quality on a mass and on microcalcifications. Also, while this is a small sample size, this preliminary study does suggest that further investigation of the scatter correction algorithm's ability to improve the image quality of clinical images is warranted. In the future, we want to determine the effects of the proposed method on more clinical DBT acquisitions that contain lesions and microcalcifications as well as images depicting other typical mammographic findings such as architectural distortions. Finally, the impact of the algorithm on actual clinical performance needs to be evaluated. We will also work toward bringing the proposed method to the clinical realm by developing either an extensive library of SPR maps of breasts of varying sizes and shapes or a computational model of SPR maps that can be used to quickly produce a patient-matched SPR map. The proposed algorithm, not including MC simulation time, can be applied in a matter of seconds, which would allow for a seamless inclusion of this technique in clinical DBT. In addition, further work on the noise filtering would include extensively testing multiple variance levels and kernel sizes for use in the adaptive means filter. Our parameters for the noise filtration were chosen once a noticeable improvement in noise level, with

FIG. 9. (a) ROI of uncorrected central slice of a mass in patient A image. (b) Identical ROI of the same slice, following scatter correction of the projections. These images are displayed with equal window width but different window levels.

FIG. 10. (a) ROI of uncorrected central slice of a microcalcification cluster in patient B image. White arrow points to the smallest lesion seen in the entire image stack. (b) Identical ROI of the same slice of (a), following scatter correction of the projections. White arrow points to the smallest lesion, more visible after the correction. (c) ROI of uncorrected central slice of a large microcalcification of patient B in a focal plane separate from (a). (d) Identical ROI of the same slice of (c), following scatter correction of the projections. These images are displayed with equal window width but different window levels.

minimal impact on sharpness, was observed. Other noise filters might also offer further improvement.

V. CONCLUSION

The software-based scatter correction algorithm proposed here was successfully implemented and tested on digital breast tomosynthesis acquisitions of both phantoms and patients. We found measurable improvements in the image quality of the scatter corrected reconstructions of phantoms of two sizes. The application of the proposed method to the reconstructions of the patient images also improved the image quality, including that of actionable findings in a clinical setting. This study has demonstrated the feasibility of a software approach to scatter reduction in DBT and could lead to both improved clinical performance of DBT and faster widespread adoption of this technology.

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APPENDIX: DERIVATION OF THE SCATTER ESTIMATE

Here we present the derivation for the equation that defines the estimated scatter signal in a DBT acquired projection (S_e). The total signal of an acquired projection (T_{acq}) consists of the scatter (S) and primary (P) signal:

$$T_{acq} = S + P \quad (A1)$$

We also define the scatter-to-primary ratio (SPR) on a pixel by pixel basis:

$$SPR = \frac{S}{P} \quad (A2)$$

Dividing equation A1 by S , we obtain:

$$\frac{T_{acq}}{S} = 1 + \frac{P}{S} = 1 + \frac{1}{SPR} \quad (A3)$$

Rearranging equation A3 results in:

$$S = T \left(\frac{SPR}{1 + SPR} \right) \quad (A4)$$

Our MC simulations generate the estimated total (T_e) image and the estimated SPR map (SPR_e). We deform and register T_e to T , as described in the Materials and Methods section, to obtain the image transformation R :

$$T = R\{T_e\} \quad (A5)$$

We then apply R to SPR_e to obtain the registered SPR map ($R\{SPR_e\}$), and low pass filter T_{acq} to obtain T_{acq-LP} . Thus S_e can be calculated as follows:

$$S_e = T_{acq-LP} \left(\frac{R\{SPR_e\}}{1 + R\{SPR_e\}} \right) \quad (A6)$$

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- ¹L. T. Niklason *et al.*, "Digital tomosynthesis in breast imaging," *Radiology* **205**, 399–406 (1997).
 - ²T. Wu, J. Zhang, R. Moore, E. Rafferty, D. Kopans, W. Meleis, and D. Kaeli, "Digital tomosynthesis mammography using a parallel maximum-likelihood reconstruction method," *Proc. SPIE* **5368**, 1–11 (2004).
 - ³I. Sechopoulos, S. Suryanarayanan, S. Vedantham, C. D'Orsi, and A. Karellas, "Computation of the glandular radiation dose in digital tomosynthesis of the breast," *Med. Phys.* **34**, 221–232 (2007).
 - ⁴R. H. Moore, D. B. Kopans, E. A. Rafferty, D. Georgian-Smith, R. A. Hitt, and E. D. Yeh, "Initial callback rates for conventional and digital breast tomosynthesis mammography comparison in the screening setting," in *Radiological Society of North America 93rd Scientific Assembly and Annual Meeting Program* (Radiological Society of North America, Chicago, IL, 2007), p. 381.
 - ⁵M. A. Helvie, M. A. Roubidoux, L. M. Hadjiiski, Y. Zhang, P. L. Carson, and H.-P. Chan, "Tomosynthesis mammography versus conventional mammography: Comparison of breast masses detection and characterization," in *Radiological Society of North America 93rd Scientific Assembly and Annual Meeting Program* (Radiological Society of North America, Chicago, IL, 2007), p. 381.
 - ⁶S. P. Poplack, T. D. Tosteson, C. A. Kogel, and H. M. Nagy, "Digital breast tomosynthesis: Initial experience in 98 women with abnormal digital screening mammography," *Am. J. Roentgenol.* **189**, 616–623 (2007).
 - ⁷I. Andersson, D. Ikeda, S. Zackrisson, M. Ruschin, T. Svahn, P. Timberg, and A. Tingberg, "Breast tomosynthesis and digital mammography: A comparison of breast cancer visibility and BIRADS classification in a population of cancers with subtle mammographic findings," *Eur. Radiol.* **18**, 2817–2825 (2008).
 - ⁸I. Sechopoulos, S. Suryanarayanan, S. Vedantham, C. J. D'Orsi, and A. Karellas, "Scatter radiation in digital tomosynthesis of the breast," *Med. Phys.* **34**, 564–576 (2007).
 - ⁹G. Wu, J. G. Mainprize, J. M. Boone, and M. J. Yaffe, "Evaluation of scatter effects on image quality for breast tomosynthesis," *Proc. SPIE* **6510**, 65101T–65109T (2007).
 - ¹⁰G. Wu, J. G. Mainprize, J. M. Boone, and M. J. Yaffe, "Evaluation of scatter effects on image quality for breast tomosynthesis," *Med. Phys.* **36**, 4425–4432 (2009).
 - ¹¹B. Liu, T. Wu, R. H. Moore, and D. B. Kopans, "Monte Carlo simulation of x-ray scatter based on patient model from digital breast tomosynthesis," *Proc. SPIE* **6142**, 61421N–61429N (2006).
 - ¹²S. Agostinelli *et al.*, "Geant4—A simulation toolkit," *Nucl. Instrum. Methods A* **506**, 250–303 (2003).
 - ¹³J. Allison *et al.*, "Geant4 developments and applications," *IEEE Trans. Nucl. Sci.* **53**, 270–278 (2006).
 - ¹⁴I. Sechopoulos and C. J. D'Orsi, "Glandular radiation dose in tomosynthesis of the breast using tungsten targets," *J. Appl. Clin. Med. Phys.* **9**, 161–171 (2008).
 - ¹⁵B. Ren, C. Ruth, T. Wu, Y. Zhang, A. Smith, L. Niklason, C. Williams, E. Ingall, B. Polischuk, and Z. Jing, "A new generation FFDM/tomosynthesis fusion system with selenium detector," *Proc. SPIE* **7622**, 76220B–76211B (2010).
 - ¹⁶J. M. Boone, T. R. Fewell, and R. J. Jennings, "Molybdenum, rhodium, and tungsten anode spectral models using interpolating polynomials with application to mammography," *Med. Phys.* **24**, 1863–1874 (1997).
 - ¹⁷A. P. Colijn and F. J. Beekman, "Accelerated simulation of cone beam x-ray scatter projections," *IEEE Trans. Med. Imaging* **23**, 584–590 (2004).
 - ¹⁸R. L. Siddon, "Fast calculation of the exact radiological path for a three-dimensional CT array," *Med. Phys.* **12**, 252–255 (1985).
 - ¹⁹J. M. Boone and A. E. Chavez, "Comparison of x-ray cross sections for diagnostic and therapeutic medical physics," *Med. Phys.* **23**, 1997–2005 (1996).
 - ²⁰R. C. Gonzalez and R. E. Woods, *Digital Image Processing*, 2nd ed. (Prentice Hall, Upper Saddle River, NJ, 2002).
 - ²¹T. Pavlidis, *Algorithms for Graphics and Image Processing* (Computer Science Press, Rockville, MD, 1982).
 - ²²R. Chandrasekhar and Y. Attikiouzel, "A simple method for automatically locating the nipple on mammograms," *IEEE Trans. Med. Imaging* **16**, 483–494 (1997).
 - ²³F. L. Bookstein, "Principal warps: Thin-plate splines and the decomposition of deformations," *IEEE Trans. Pattern Anal. Mach. Intell.* **11**, 567–585 (1989).
 - ²⁴R. C. Gonzalez and R. E. Woods, *Digital Image Processing*, 3rd ed. (Prentice Hall, Upper Saddle River, NJ, 2008).
 - ²⁵T. Wu, R. H. Moore, E. A. Rafferty, and D. B. Kopans, "A comparison of reconstruction algorithms for breast tomosynthesis," *Med. Phys.* **31**, 2636–2647 (2004).
 - ²⁶I. S. Kyprianou, A. Badano, B. D. Gallas, and K. J. Myers, "A method to estimate the point response function of digital x-ray detectors from edge measurements," *Proc. SPIE* **6510**, 65100B–65112B (2007).
 - ²⁷H. Fujita, D. Y. Tsai, T. Itoh, K. Doi, J. Morishita, K. Ueda, and A. Ohtsuka, "A simple method for determining the modulation transfer function in digital radiography," *IEEE Trans. Med. Imaging* **11**, 34–39 (1992).
 - ²⁸D. Förmvik, I. Andersson, T. Svahn, P. Timberg, S. Zackrisson, and A. Tingberg, "The effect of reduced breast compression in breast tomosynthesis: Human observer study using clinical cases," *Radiat. Prot. Dosim.* **139**, 118–123 (2010).
 - ²⁹R. S. Saunders, E. Samei, J. Y. Lo, and J. A. Baker, "Can compression be reduced for breast tomosynthesis? Monte Carlo study on mass and microcalcification conspicuity in tomosynthesis," *Radiology* **251**, 673–682 (2009).