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A scatter correction method for contrast-enhanced dual-energy digital breast tomosynthesis

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Abstract

Contrast-enhanced dual energy digital breast tomosynthesis (CE-DE-DBT) is designed to image iodinated masses while suppressing breast anatomical background. Scatter is a problem, especially for high energy acquisition, in that it causes severe cupping artifact and iodine quantitation errors. We propose a patient specific scatter correction (SC) algorithm for CE-DE-DBT. The empirical algorithm works by interpolating scatter data outside the breast shadow into an estimate within the breast shadow. The interpolated estimate is further improved by operations that use an easily obtainable (from phantoms) table of scatter-to-primary-ratios (SPR) - a single SPR value for each breast thickness and acquisition angle. We validated our SC algorithm for two breast emulating phantoms by comparing SPR from our SC algorithm to that measured using a beam-passing pinhole array plate. The error in our SC computed SPR, averaged over acquisition angle and image location, was about 5%, with slightly worse errors for thicker phantoms. The SC projection data, reconstructed using OS-SART, showed a large degree of decupping. We also observed that SC removed the dependence of iodine quantitation on phantom thickness. We applied the SC algorithm to a CE-DE-mammographic patient image with a biopsy confirmed tumor at the breast periphery. In the image without SC, the contrast enhanced tumor was masked by the cupping artifact. With our SC, the tumor was easily visible. An interpolation-based SC was proposed by (Siewerdsen *et al.*, 2006) for cone-beam CT (CBCT), but our algorithm and application differ in several respects. Other relevant SC techniques include Monte-Carlo and convolution-based methods for CBCT, storage of a precomputed library of scatter maps for DBT, and patient acquisition with a beam-passing pinhole array for breast CT. Our SC algorithm can be accomplished in clinically acceptable times, requires no additional imaging hardware or extra patient dose and is easily transportable between sites.

1. Introduction

Conventional X-ray mammography and digital breast tomosynthesis (DBT) rely on attenuation differences to show masses. Masses in conventional mammographic images may be masked by over- and underlying tissue structure, but even in DBT, a mass might not have sufficient contrast to be easily visualized. In this case one can use iodine contrast-enhanced (CE) X-ray imaging, which relies on the fact that, due to angiogenesis, malignant masses

contain a dense network of leaky capillaries (Dromain *et al.*, 2012). The contrast material leaks through to capillaries to form, ideally, an easily seen bolus of high-attenuation iodine at the location of the mass (Dromain *et al.*, 2012) (Jong *et al.*, 2003).

Two basic methods for CE breast imaging, temporal subtraction (Skarpathiotakis *et al.*, 2002) and dual-energy (DE) imaging (Puong *et al.*, 2007), have been proposed. We focus on DE here because of its practical acquisition advantages (Puong *et al.*, 2007), and reduction in patient motion artifacts (Carton *et al.*, 2006). In CE-DE imaging both a diagnostic-range low energy (LE) acquisition, and a high energy (HE) acquisition with mean energy above K-edge of iodine, are acquired. In CE-DE digital mammography (CE-DE-DM), the HE and LE images, acquired at one angle, undergo a weighted subtraction to reveal a 2-D image of the iodine concentration with the anatomical tissue background image suppressed (Lewin *et al.*, 2003). In CE-DE-DBT, the subtracted images at each angular acquisition serve as an input to a reconstruction algorithm. The resulting 3-D reconstruction shows the iodine concentration with the anatomical background suppressed (Puong *et al.*, 2007). In addition to finding suspicious lesions CE breast imaging has also been applied to obtain contrast kinetics (Dromain *et al.*, 2006), where the kinetic behavior might be used to discriminate benign from cancerous lesions in a manner analogous to that used in Gd-contrast breast MRI.

Conventional breast tomosynthesis has the advantage of displaying a 3-D anatomical image instead of the 2-D projection image of conventional mammography. CE-DE-DBT enjoys this same advantage relative to CE-DE-DM for 3D contrast display. Neither CE-DE-DM nor CE-DE-DBT yet enjoys routine clinical use. Improvements in image quality may further more widespread use. One improvement is effective patient-specific scatter correction. In this paper, we address scatter correction for CE-DE-DBT.

Scatter diminishes iodine visualization and quantification accuracy (Carton *et al.*, 2006; Boone and Cooper III, 2000; Puong *et al.*, 2008; Salvagnini *et al.*, 2012). In particular, for HE acquisitions, scatter can create severe cupping artifacts in the projection data -far more severe than those from diagnostic LE acquisition- that propagate into the reconstruction. Also, without some means of scatter correction (SC), the HE and LE acquisition parameters and weighting factor cannot be optimally selected to suppress anatomical tissue background variations (Carton *et al.*, 2006; Boone and Cooper III, 2000; Puong *et al.*, 2008; Salvagnini *et al.*, 2012). Even for HE DBT acquisitions with no iodine contrast, scatter can significantly reduce image quality. In a DBT study in (Wu *et al.*, 2009), Monte-Carlo based techniques were used to obtain reconstructions with perfect SC (scatter counts eliminated), no SC (scatter counts retained but no correction), and with some SC. Their SC was obtained by simulating specialized grids. As more SC was introduced, lesion (mass) contrast and SDNR (signal-difference to noise ratio) improved.

In order to be useful, the SC computation must be accomplished in a clinically acceptable time. Furthermore, a successful SC technique should not add significant patient acquisitions nor result in the patient spending significant extra time under compression. In addition, a successful SC method must be transportable from one site to another, without a labor-intensive recalibration of site-specific parameters.

A variety of SC methods have been studied for DM and DBT as well as for the related modalities of cone-beam X-ray CT (CBCT) and dedicated breast CT. (Sisniega *et al.*, 2015) used GPU-accelerated Monte Carlo (MC) applied to brain CBCT to rapidly calculate patient-specific scatter estimates. This technique required an initial fully-3D (not limited angle) reconstruction, using uncorrected data followed by anatomical segmentation. MC was then applied to the segmented image to directly obtain scatter estimates. Processing times of 14 min were reported. Investigators have computed scatter point-spread functions (spsf) (Yang *et al.*, 2014; Lau, 2012; Diaz *et al.*, 2014) to study the nature of scatter in DM and DBT. (Zhao *et al.*, 2015) developed a spsf-based patient specific SC method for CBCT. An initial segmented 3D reconstruction was used to estimate primary data, which could be combined with uncorrected (primary + scatter) data to fit spsf parameters and perform SC. Processing times were about 15 min. (Yang *et al.*, 2014) used a hardware approach for SC in dedicated breast CT. In addition to ordinary projection acquisitions, extra projections utilizing a pinhole array positioned at the X-ray output were used to obtain projections of primary-only data. The two types of projections were combined to obtain a scatter estimate. In another approach used for DBT patient-specific SC (Sechopoulos *et al.*, 2007; Feng and Sechopoulos, 2011; Feng *et al.*, 2014) precomputed, using offline MC, an extensive “library” of scatter maps for a variety of breast shapes and thicknesses. This method required an extensive library of (about 10^5 entries) and complex image analysis techniques to match a given breast to a library entry to obtain the scatter estimate. An empirical method (Siewerdsen *et al.*, 2006) was applied to CBCT. Here the tails of projection data due to scatter outside the object shadow were interpolated into the shadow region to obtain a scatter estimate.

We propose a SC method intended for clinical use in CE-DE-DBT. Initial results were presented in Lu et al. (Lu *et al.*, 2014a). Our method relies on interpolating the projection image in a region outside the breast shadow into a scatter estimate within the breast shadow. This has some similarities to the method in (Siewerdsen *et al.*, 2006) mentioned above. However, that method was proposed for cone beam CT and the scatter correction problem there differs from ours in many respects.

Though empirical, our SC correction is patient specific and very fast (about 5 minutes but easily accelerated to well under a minute). No extra scans or extra patient dose is required. No extra hardware or hardware development is needed. The SC package can be easily migrated to different sites, requiring only easily obtainable site specific calibration data. Interestingly, interpolation-based empirical SC is routinely used in clinical SPECT. In SPECT “window-based” SC, scatter in a photopeak energy window is estimated by interpolation of scatter data from adjacent satellite energy windows (Ogawa *et al.*, 1991).

In section 2, we present the SC algorithm and a methodology for validating the scatter estimate. In section 3, we present the validation results for a variety of phantoms. We also apply SC to phantom acquisitions to demonstrate decoupling and the removal of iodine signal dependence on breast thickness. Anecdotal clinical results are shown in section 4. In section 5 we present a discussion of results and limitations of our method.

2 Methods

2.1. Acquisition details

Testing and validation were accomplished through phantom acquisitions. All acquisitions were done on a Siemens Mammomat Inspiration CEDET DBT prototype unit shown in Figure 1. The 300 μ m a-Se detector has 3584 by 2816 square pixels of dimension 85 μ m on a side. The X-ray focal-spot to detector distance is 655.5 mm and the center of rotation is located 47 mm above the detector, with air gap between the carbon fiber detector cover and detector of 17 mm. The compression paddle is composed of 3 mm PMMA and its area covers the detector. Images were acquired at 25 angular orientations over an angular span of $\theta_{initial} = -23^\circ$ to $\theta_{final} = +23^\circ$ in steps of $\theta = 2^\circ$. The X-ray tube motion was continuous.

We used the following parameters for dual-energy phantom acquisition that we previously developed (Hu *et al.*, 2013). The tube parameters were 28 kVp for LE and 49 kVp for HE. The particular mAs used depended on the given experiment, and we report this in section 2. We used a W/Rh anode/filter combination for LE and W/Cu for HE. The filter thickness for Rh was 50 μ m and for Cu was 237 μ m.

For CE-DE-DBT studies, we first used a CIRS BR3D (model 020) phantom. (For convenience, we shall refer to this as a “CIRS-020” phantom.) The CIRS-020 phantom comprises six 1 cm thick semicircular slabs of radius 9 cm containing a mixture of simulated adipose (50%) and glandular (50%) tissue to mimic the human breast. The adipose and glandular components are spatially mixed into a “swirl” pattern as seen in Figure 2(a). These slabs can be stacked to obtain a range 2-6 cm of breast thicknesses. One customized slab contained iodine inserts arranged in a 4 $\times$ 4 array as shown in Figure 2. As shown in Figure 2, the iodine cylinders in each column are 2, 3, 5 and 8 mm in diameter with the depth equal to the diameter. The signals are spaced by 1 cm. The tops of all 16 cylinders lie in one plane. The iodine concentration in each row is (top to bottom) 1, 2, 3 and 5 mg/ml. The thickness of our reconstructed slice was 1 mm and the slice was chosen to intersect all iodine cylinders.

Since real breasts under compression have rounded edges and the CIRS-020 has sharp edges, we verified that our scatter correction was unaffected by rounded edges through use of a CIRS tissue-equivalent phantom (model 011A)(CIRS), shown in Figure 2(c), designed to mimic a 4.5 cm thick compressed breast. The lateral dimensions are 12.5 (chest to nipple) and 18.5 cm (width). (For convenience, we shall refer to this as the “CIRS-011A” phantom.) The CIRS-011A has no iodine contrast inserts. It is of uniform 50-50 adipose-glandular composition except for linear fibers, dots (calcifications), cylinders (masses) and calibrated resolution and attenuation reference inserts, but these have no effect on our validations.

2.2. Imaging and reconstruction model for CE-DE-DBT

Figure 3 illustrates a stylized (not to scale) DBT system with a stationary detector and an X-ray tube that rotates (indicated by angle θ) in the x-z plane with the y coordinate perpendicular to the paper. The system is shown at two angular orientations. Profile plots of detector readout along x at a given y are shown for each of the two angular orientations.

Consider the high-energy case with $I_{HE}^{\theta}(x, y)$ the detector readout at (x, y) . The subscript HE indicates high energy acquisitions and superscript θ indicates the acquisition angle. We can write the HE acquisition as

$$I_{HE}^{\theta}(x, y) = b_{HE}^{\theta}(x, y) \exp\left(-\int_{l_{xy}^{\theta}} \mu_{HE}(x, y, z) dl\right) + s_{HE}^{\theta}(x, y) + n_{HE}^{\theta}(x, y), \quad (1)$$

where b_{HE}^{θ} is the high-energy blank scan, s_{HE}^{θ} is high-energy scatter and n_{HE}^{θ} indicates all radiation and detector noise sources from a high energy acquisition, here written as additive term. The beam attenuation is modeled by the exponentiated line integral, where $\mu_{HE}(x, y, z)$ is the high energy linear attenuation coefficient of the breast, the compression paddle and breast support; dl indicates line integral summation with l_{xy}^{θ} denoting that the line integral is along a ray from the X-ray focal spot (here considered a point source) to the pixel centered at (x, y) . Note that Eq.(1) holds for a monoenergetic case, so it can be considered an approximation with HE representing the mean of the HE energy range. Equation (1) is written in a continuous (x, y, z) space, but it is understood that (x, y) indicates the center of an integrating detector element centered at (x, y) while for the attenuation object $\mu(x, y, z)$, the coordinates are continuous. Everything follows analogously for the low energy case, where the imaging equation becomes

$$I_{LE}^{\theta}(x, y) = b_{LE}^{\theta}(x, y) \exp\left(-\int_{l_{xy}^{\theta}} \mu_{LE}(x, y, z) dl\right) + s_{LE}^{\theta}(x, y) + n_{LE}^{\theta}(x, y) \quad (2)$$

with the subscript LE indicating low energy.

Consider the problem of reconstructing the HE data. Typically, a log step of the form

$$g_{HE}^{\theta}(x, y) = -\log\left(\frac{I_{HE}^{\theta}(x, y)}{b_{HE}^{\theta}(x, y)}\right) \quad (3)$$

is first applied. If scatter and noise are absent in Eq.(1), then $g_{HE}^{\theta}(x, y)$ is simply the line integral data of the object $\mu_{HE}(x, y, z)$. For reasons explained below, we will use the set of $g_{HE}^{\theta}(x, y)$ images (one per θ) as input to a reconstruction algorithm to obtain a 3-D HE reconstruction. However, if scatter is present in eq.(1), the detector data $I_{HE}^{\theta}(x, y)$ will contain a “cupping” profile easily seen in Figure 3. The cupping is propagated into $g_{HE}^{\theta}(x, y)$ via the log step in eq.(3) and is further propagated into the reconstruction such that a reconstructed slice shows a cupping profile. Thus if one could estimate $s_{HE}^{\theta}(x, y)$ and subtract it from $I_{HE}^{\theta}(x, y)$, the subsequent reconstruction would be “decupped” yielding better iodine image quality.

For LE data, the cupping in $I_{LE}^{\theta}(x, y)$ is negligible relative to that in $I_{HE}^{\theta}(x, y)$. Nevertheless, the scatter $s_{LE}^{\theta}(x, y)$, when propagated through the log step and into an LE reconstruction, causes quantitative errors. Thus estimation of $s_{LE}^{\theta}(x, y)$ and subtraction from eq. (2) yields a quantitatively improved reconstruction.

Above we discussed separate reconstructions of the HE and LE data, which differs from the conventional strategy of weighted subtraction-then-reconstruction strategy. Conventionally, in CE-DE-DBT, a weighted subtraction of the form

$$\Delta g^{\theta}(x, y) = g_{HE}^{\theta}(x, y) - W \cdot g_{LE}^{\theta}(x, y) \quad (4)$$

is performed, where $g^{\theta}(x, y)$ is weighted subtracted projection data at θ and W is a weight factor. This yields $g^{\theta}(x, y)$, a projection image at θ , where ideally, the contrast between adipose and glandular tissue is zero, leaving only an iodine projection image (Samei and Saunders, 2011). The $g^{\theta}(x, y)$ can then be reconstructed to ideally show iodine contrast only. In practice, polyenergetic effects and patient motion lead to an incomplete subtraction of the breast structure which leads to residual anatomical structure in the CE-DE-DBT reconstruction. The noise will propagate from the projection data to yield correlated noise in the CE-DE-DBT reconstruction. Scatter results in the “cupping” artifact, mainly due to cupping in HE data, that can greatly reduce visibility of the iodine signal in the CE-DE-DBT reconstruction.

We modify the conventional strategy in (4) by separately reconstructing g_{HE}^{θ} and g_{LE}^{θ} and then performing the weighted subtraction in the reconstruction domain (Chen *et al.*, 2013). The reconstruction algorithm used in this study was OS-SART (Wang and Jiang, 2004). We used 4 iterations. (We avoided the use of filtered backprojection algorithms since they yield artifacts (Lu *et al.*, 2014b; Chen *et al.*, 2013) in the reconstruction of the iodine signals.) Since OS-SART is linear, it can be shown that the choice of weight is unaffected. The advantage (Chen *et al.*, 2013) of our reconstruct-then-subtract strategy is that it minimizes artifacts due to inexact angular positioning of the X-ray tube. Scatter estimates were subtracted from the projection data (before the log step) before the reconstruction.

2.3. Scatter correction algorithm

2.3.1 Overview—With the aid of Figure 3, we illustrate the segmentation of each HE detector readout into three regions. Procedures for the segmentation are given in section 2.3.2. The segmentation is illustrated along x for a given value of y . The “wing” region is delineated by the X-ray tangent to the breast. In this region, the readout comprises scatter and blank scan. It is easy to subtract the known blank scan b_{HE}^{θ} in the wing region to obtain s_{HE}^{θ} in the wing region, shown as the yellow area in Figure 3. The “object” region comprises that part of the detector readout that has received rays passing through the full thickness of the compressed breast. The “penumbra” region is the remaining segment in between the wing and object regions.

We first give a simple and intuitive overview of the HE SC algorithm with the aid of the scatter profiles in Figure 4. In section 2.3.2, we revisit the SC algorithm in mathematical detail. Note that the LE SC algorithm is a simplified version of the HE SC and will be described in section 2.3.2.

In Figure 4(a), the true scatter profile is shown in the 3 regions. As can be seen in Figure 3, the scatter in the wing region can be obtained by subtraction of the (known) blank scan. Our goal is to obtain an estimate of the true scatter in the object region using knowledge of the wing data along with pre-stored information.

We presume that the profile in the object region is a flattened bell-shaped curve. As a first step, we use polynomial-based interpolation of the wing data to obtain an initial estimate of the scatter in the object and penumbral regions. The interpolated data is shown in Figure 4(b). As we will validate, with an appropriate choice of the interpolating polynomial, the shape of the interpolated curve in the object region is nearly correct and differs from the true scatter data by an additive constant k . This feature is illustrated in Figure 4 (c). We will be able to obtain k using a relatively simple procedure involving pre-stored measurements as detailed in section 2.3.2.

We add k to the interpolated scatter estimate in the object and penumbral regions as seen in Figure 4(d). In Figure 4(d) we see that resulting scatter estimate is fairly accurate in the object region but is a poor approximation of the true scatter in the penumbral region. The poor approximation of scatter in the penumbral region causes artifacts in the dual-energy reconstruction as will be shown in later sections, but these artifacts are insignificant compared to artifacts due to not doing any scatter correction in the first place. The scatter estimate for this simple 1-D case is shown in Figure 4(d).

In Siewerdsen et al. (Siewerdsen *et al.*, 2006), the notion of scatter estimation by interpolated wing data is used in the context of cone-beam CT. The interpolation in itself is supplemented by an additional empirical step involving truncation of the wing region before interpolation.

2.3.2 Details—We describe the SC algorithm in detail with the aid of the table of symbols in Table 1 and the pseudocode in Table 2. We first consider HE data only. The SC operations are performed at each angle, as indicated by the superscript θ . For a given patient, the HE projection data $I_{HE}^{\theta}(x, y)$ is acquired along with a pre-measured and dose-scaled blank scan $b_{HE}^{\theta}(x, y)$.

We first list the steps used in segmenting and extracting the wing data $w_{HE}^{\theta}(x, y)$ from $I_{HE}^{\theta}(x, y)$ and $b_{HE}^{\theta}(x, y)$. We convolve the blank scan $b_{HE}^{\theta}(x, y)$ with a Gaussian kernel (limited to 15×15 pixel extent and with $\sigma = 7$ pixels) to obtain a smoothed blank scan $\tilde{b}_{HE}^{\theta}(x, y)$. Wing data $w_{HE}^{\theta}(x, y)$ is obtained by the following operation:

$$w_{HE}^{\theta}(x, y) = [(I_{HE}^{\theta}(x, y) - \tilde{b}_{HE}^{\theta}(x, y)) \otimes g(x, y)]_{+} \quad (5)$$

The subtraction in (5) yields wing data which is further smoothed by convolution ($\otimes$) with a Gaussian kernel $g(x, y)$ restricted to a 5×5 pixel region and with $\sigma = 3$. Any negative values are set to zero as indicated by the clipping operation $[\]_{+}$. The subtraction and clipping operations perform the essential segmentation step for wing data since the penumbral and object region data lie below the blank scan as depicted in Figure 3. The cyan pixels in Figure 5 show an x-profile of $w_{HE}^{\theta}(x, y)$. The segmentation is not perfect; as seen in Figure 5 (a), there are a few outlier pixels on the inner edges of the HE wing data. These are due to a region in the penumbra adjacent to the wing, where the penumbral values exceed the blank scan. Figure 3 illustrates this penumbral effect.

We obtain the interpolated HE scatter profile of Figure 4(b) by performing, for each value of y , a 1-D polynomial interpolation between the two wing regions. Our validation studies in section 3.1 show that a polynomial of order 4 yields the best results. To implement the 4th order 1-D polynomial interpolation, we used the Matlab 2013a curve fitting function *fit()*. Note that the effect of the outlier pixels on the interpolation can be controlled by selecting a robust interpolation option (LAR) in Matlab. (We could have modified the wing segmentation to eliminate the outlier pixels but we found that robust interpolation was easier and led to equally good results.) The green curve in Figure 5(a) shows the resulting interpolated scatter estimate. The interpolation along x for a fixed y is repeated for each y to obtain a surface $\tilde{s}_{HE}^{\theta}(x, y)$ smooth in x but choppy in y as seen in Figure 5(b). The “choppiness” stems from the fact that each x interpolation is from a different noisy set (different y) of wing data. Since we know that the scatter field should be smooth in y , we apply a further smoothing in the y direction using a 1-D Gaussian kernel ($\sigma = 60, 200$ pixel extent) to yield a smooth surface $\hat{s}_{HE}^{\theta}(x, y)$ as illustrated in Figure 5(c).

We assume (and later validate) that $\hat{s}_{HE}^{\theta}(x, y)$ will have an approximately correct shape within the object region, but will be quantitatively inaccurate. It underestimates the true scatter $s_{HE}^{\theta}(x, y)$ in the object region by an additive *constant* k_{HE}^{θ} as depicted in Figure 4(c), so that

$$k_{HE}^{\theta} = s_{HE}^{\theta}(x, y) - \hat{s}_{HE}^{\theta}(x, y). \quad (6)$$

(Note that k depends on θ and energy.) This underestimate is due to the fact that we cannot accurately interpolate through the rapidly rising scatter in the penumbral region. Figure 4(d) depicts this underestimation.

A key step in the SC algorithm is determination of k_{HE}^{θ} . If we assume that eq.(6) holds for all the points in the object region, then it must hold at a reference location $(x_c^{\theta}, y_c^{\theta})$ in the object region:

$$k_{HE}^{\theta} = s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) - \hat{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}), \quad (7)$$

where $(x_c^{\theta}, y_c^{\theta}) \equiv \arg \max_{x, y} [\hat{s}_{HE}^{\theta}(x, y)]$. The true scatter $s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ is unknown, but we can approximate it with aid of the scatter-to-primary ratio evaluated at $(x_c^{\theta}, y_c^{\theta})$:

$$SPR_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) = \frac{s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})}{p_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})} \equiv \frac{s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})}{I_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) - s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})}, \quad (8)$$

where $p_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ is the intensity of the primary (unscattered) rays. Solving eq.(8) for $s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ yields

$$s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) = \left[\frac{SPR_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})}{1 + SPR_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})} \right] I_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}). \quad (9)$$

where $I_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ is just the patient measurement data at point $(x_c^{\theta}, y_c^{\theta})$.

We do not know $SPR_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ for a given patient but we can approximate it using a library of precomputed point measurements obtained from breast-equivalent phantoms as detailed in section 2.4. In general, the scatter-to-primary ratio at a central point depends (Sechopoulos *et al.*, 2007) on thickness t , energy, angle θ and to a lesser extent the glandular portion. The scatter-to-primary ratio measured at $(x_c^{\theta}, y_c^{\theta})$ for the patient breast will be close in value to the scatter-to-primary ratio measured at a suitably defined central point of a phantom $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$. For a breast and its phantom equivalent, i.e. the phantom that matches the breast in t and a , we can thus write

$$SPR_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) \approx \overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta}) \quad (10)$$

where the bar superscript indicates a phantom quantity. Thus our problem is one of determining from phantoms a library of scatter-to-primary *point* measurements, not a library of scatter-to-primary *images*. In section 2.4, we detail a procedure for determining $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ and $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ as well as the same quantities for the LE case.

From eqs.(9) and (10), we may write

$$\hat{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) = \left[\frac{\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})}{1 + \overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})} \right] I_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}), \quad (11)$$

where $\hat{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ is an estimate of $s_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$ due to the approximation in Eq.(10). We use eqs. (7) and (11) to obtain

$$k_{HE}^{\theta} \approx \hat{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) - \tilde{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}). \quad (12)$$

From eq.(6), eq.(12) holds at all (x, y) in the object region, so that

$$\hat{s}_{HE}^{\theta}(x, y) \approx \tilde{s}_{HE}^{\theta}(x, y) + k_{HE}^{\theta}. \quad (13)$$

where $\hat{s}_{HE}^{\theta}(x, y)$ is our final scatter estimate. This estimate can be subtracted from eq.(1) to perform scatter correction.

To obtain the low energy scatter estimate in the object region $\hat{s}_{LE}^{\theta}(x, y)$, we follow the steps for the HE case but set $\tilde{s}_{LE}^{\theta}(x, y) = 0$. The reason for this is that the wing data for the LE case is essentially zero due to the narrowness of the LE spsf. Within the object region we approximate $\hat{s}_{LE}^{\theta}(x, y)$ as a constant k_{LE}^{θ} . Numerous studies (Sechopoulos *et al.*, 2007; Feng *et al.*, 2014; Boone and Cooper III, 2000) have shown a slow spatial variation of low energy scatter in the object region, but we use a constant because the LE spatial variation is so much smaller than the HE scatter variation that the constant approximation yields good results. The remaining steps follow as in the HE case. In particular, we build a library $\overline{SPR}_{LE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ using LE acquisitions of phantoms of varying thicknesses and use these values in eq.(11).

One might suspect that we could apply this constant scatter correction strategy for HE as well, setting $\hat{s}_{HE}^{\theta}(x, y) = k_{HE}^{\theta}$. If degradation in the scatter correction performance using the constant scatter subtraction for HE and LE is not too bad, then we might elect to use this strategy, thus avoiding the extra effort in implementing polynomial methods. In our evaluations below we test this strategy.

The SC algorithm for HE is summarized in the pseudocode of Table 2. The LE SC algorithm, described above, is a simplification of the HE algorithm. For convenience, the term $G_{15 \times 15}^{\sigma=7}(x, y)$ refers to a 2-D Gaussian smoothing kernel with standard deviation 7 pixels and extent limited to 15×15 pixels. This construct also appears in the kernels $G_{5 \times 5}^{\sigma=3}(x, y)$ and the 1-D kernel $G_{1 \times 200}^{\sigma=60}(y)$. Figure 6 illustrates the x-profiles of all processing stages of the algorithm for HE acquisitions using a CIRS-020 phantom at $\theta = 0^{\circ}$ and $\theta = 23^{\circ}$ in place of a real breast.

2.4. Determining pre-measured library quantities

From section 2.3.2, the SC algorithm requires knowledge of $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ and $\overline{SPR}_{LE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$, the scatter-to-primary ratios measured from reference phantoms. Again, we discuss measurements in terms of HE, but this will apply equally to the LE case. The quantities $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ comprise a library of scalars indexed by θ and phantom thickness t . (In the notation for the library quantity $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ we suppress t .) For a given patient,

we use the measured compressed breast thickness to find the corresponding value of t . The thickness t should ideally equal the thickness of compressed patient breast. For our own work, we have available phantoms of thicknesses 2, 3, 4, 5 and 6 cm obtained by stacking 1 cm thick CIRS-020 slabs. We interpolate the $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ values to match the thickness of the patient breast. We note that scatter-to-primary ratio also varies with the glandular/adipose composition ratio, but this variation is slow (Sechopoulos *et al.*, 2007). The glandular/adipose ratio for the CIRS-020 is fixed at 50%, but given the weak dependence of the scatter-to-primary ratio on this compositional ratio, we do not incur a significant error in $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$.

We will need to measure a scatter-to-primary ratio measurement only in the central region about $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ of the phantom corresponding to a central region about $(x_c^{\theta}, y_c^{\theta})$ of the compressed breast. Since the scatter-to-primary ratio at a central location is insensitive to the overall breast shape (Sechopoulos *et al.*, 2007), the shape of the phantom need not match the shape of the patient breast. The point $(x_c^{\theta}, y_c^{\theta})$ is the location where $\hat{s}_{HE}^{\theta}(x, y)$ attains a maximum, which in turn is the location where $\hat{s}_{HE}^{\theta}(x, y)$ attains a maximum by virtue of eq. (13). Since $\hat{s}_{HE}^{\theta}(x, y)$ is approximately equal to $s_{HE}^{\theta}(x, y)$, $(x_c^{\theta}, y_c^{\theta})$ is the approximate location of maximum scatter in the patient data. We therefore measure $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ of the reference phantom at a point $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ where scatter is a maximum. To find this maximum, we must measure the phantom scatter-to-primary ratio at an array of locations. We can obtain these measurements using techniques that entail the use of a pinhole array (Chen *et al.*, 2012; Inscoe *et al.*, 2013; Yang *et al.*, 2014). Details of our adaptation and use of the pinhole array measurement technique (Chen *et al.*, 2012; Inscoe *et al.*, 2013; Yang *et al.*, 2014) are found in section 2.5 as part of our description of validation techniques. We define, for a given θ , $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ as that pinhole array shadow location that is within the object region of the breast shadow and also has the maximum scatter-to-primary ratio. This scatter-to-primary ratio value is our $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$. By performing the procedure at HE and LE, we obtain $\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ and $\overline{SPR}_{LE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$ and use these quantities in eq.(11).

2.5. Pinhole array technique for scatter-to-primary ratio measurement

The pinhole array method can deliver measurements of primary-only photons for a given phantom. Below we describe our use of this technique and also cross-check its ability to deliver primary-only measurements with the aid of a beam-blocker technique (Segui and Zhao, 2006; Niu and Zhu, 2011; Lazos and Williamson, 2012). We describe the technique for the HE case only, but the steps are the same for the LE case.

We place a lead plate with a pinhole atop a 3.9 cm thick uniform Lucite phantom. The plate thickness is 3 mm and hole diameter 1 mm. The lead plate mounted on the scanner is tilted by θ so that it is perpendicular to the X-ray beam. At any θ the plate is positioned so the focal-spot-to-hole distance remains constant. The pinhole shadow will comprise only primary photons as we will verify below. The plate is then removed and the exposure repeated. Now the photons in the pinhole shadow region comprise total (scatter + primary)

photons. From the pinhole shadow, we obtain a measure of the primary photons p by averaging a 5×5 pixel region at the center of the approximately circular shadow. With the plate removed, we can obtain a measure of total counts $I = p + s$ by averaging over the same region. The two quantities I and p can be subtracted to obtain the scatter s . Thus from the two acquisitions, we can obtain the scatter s and the primary p for the phantom at the pinhole shadow. From the two measurements, we can thus compute the true scatter-to-primary ratio $SPR = s / p$ within the pinhole shadow region. The acquisitions can be done at non-clinical high mAs to obtain nearly noiseless quantities. We can obtain SPR at all θ by tilting the plate. The location of the pinhole shadow will vary with θ . The HE acquisitions with and without the plate are acquired at 36 mAs per angle and for the LE case 54 mAs per angle.

To verify that for the given pinhole dimensions the reading in the shadow region comprises only primary photons, we performed the following cross-check. We used a beam-blocker method (Segui and Zhao, 2006; Niu and Zhu, 2011; Lazos and Williamson, 2012) with blocking disks of 2mm - 7 cm for the case of $\theta = 0^\circ$ and for HE and LE exposures equal to that used for the single-pinhole case. The beam blocker method will deliver a scatter value s_{BB} at the pinhole shadow location. The pinhole method delivers $I - p = s$. Our cross-check involved comparing s_{BB} with s . The difference with the two methods was less than 0.5%. In order to reduce discrepancies in s_{BB} vs. s due to phantom inhomogeneity, we used the uniform Lucite phantom. With the uniform phantom, exact positioning of the pinhole versus the blocking disks was not critical.

The single-pinhole experiment above was used to verify that we could obtain s and p in the pinhole shadow. However, to determine $(\bar{x}_c^\theta, \bar{y}_c^\theta)$ and $\overline{SPR}_{HE}^\theta(\bar{x}_c^\theta, \bar{y}_c^\theta)$ we need an array of pinholes. We use a 3 mm lead plate oriented at θ as before atop the Lucite phantom, but now the lead plate contains an 15×15 array of pinholes arranged in a square grid with hole centers spaced at 1 cm. Figure 7 shows the pinhole placement along with a picture of the lead pinhole plate. The HE and LE acquisition parameters were the same as for the single-pinhole case.

The photons received in each of the 225 shadow regions constitute primary photons $p_{HE}^\theta(x, y)$. With the plate removed, we receive photons $I_{HE}^\theta(x, y) = p_{HE}^\theta(x, y) + s_{HE}^\theta(x, y)$ in the 225 shadow regions; by subtraction, we obtain $s_{HE}^\theta(x, y)$ separately. Within each shadow region, we average a 5×5 pixel region at the center of the shadow to obtain scalars $s_{HE}^\theta(x, y)$ and $p_{HE}^\theta(x, y)$ where (x, y) is now understood to be the pixel location at the center of the

shadow. Then we may set $SPR_{HE}^\theta(x, y) \equiv \frac{s_{HE}^\theta(x, y)}{p_{HE}^\theta(x, y)}$ as the scatter-to-primary ratio at each pinhole shadow. We use the same procedure for LE to obtain $SPR_{LE}^\theta(x, y)$. Figure 8 shows a reading with plate present and absent over a region that includes three pinhole shadows.

The pinhole-array method requires that contributions through one pinhole do not add counts to other pinhole shadows. To verify that pinholes do not interact, we use the Lucite phantom and acquire acquisitions with plate present and absent as described previously. We measure

contributions at θ for all 225 holes and, by using a second plate to cover all holes except one reference hole, measure the contribution due to the reference hole. Figure 9 illustrates superposed profiles from the two measurements. Qualitatively, it is seen in Figure 9 that the holes do not interact. For a quantitative comparison, we computed the mean m_1 of a 5×5 pixel region in the reference hole shadow with all other holes blocked and a similar mean m_2 of the pixels in the reference hole shadow with all other holes open. We found that

$\frac{m_2 - m_1}{m_1} < 0.01$. Note that m_1 and m_2 are essentially deterministic due to the very high dose of the measurements. i.e. the standard deviations of m_1 and m_2 from repeated measurements are less than $m_2 - m_1$. We repeated this procedure for two more reference holes and obtained similar results.

Our application of the pinhole array technique extends the work in (Chen *et al.*, 2012; Inscoe *et al.*, 2013) in two ways. First, we used a beam blocker method to cross-check the primary readings as describe above, and we also applied this technique in a geometry where the X-ray tube and pinhole plate were positioned at angle θ relative to the detector.

Exposures with and without a pinhole array can be used to directly compute scatter at dense array of points in a given projection. Thus one might simply mount a pinhole plate on the X-ray output during clinical exposure and compute scatter without going through the many empirical steps we propose. Indeed, for dedicated breast-CT, (Yang *et al.*, 2014) proposed exactly this technique, though the number of angles for which the pinhole plate was used is limited. In our work, we avoid the strategy of directly using a pinhole plate since it entails extra patient exposure and also requires hardware development for each new DBT site. However, we do indeed use the pinhole plate for validation purposes as described in the next section.

2.6. Validation of scatter correction algorithm

We describe our validation for the HE case; the LE case follows. In section 2.5, we showed how we can use pinhole array techniques to obtain ground-truth values $SPR_{HE}^{\theta}(x, y)$ at the pinhole shadow locations. At these locations, we compare $SPR_{HE}^{\theta}(x, y)$ to $\hat{SPR}_{HE}^{\theta}(x, y)$, where $\hat{SPR}_{HE}^{\theta}(x, y)$ is obtained using the scatter estimate $\hat{s}_{HE}^{\theta}(x, y)$ of eq.(13) as follows:

$$\hat{SPR}_{HE}^{\theta}(x, y) \equiv \frac{\hat{s}_{HE}^{\theta}(x, y)}{\hat{p}_{HE}^{\theta}(x, y)} = \frac{\hat{s}_{HE}^{\theta}(x, y)}{I_{HE}^{\theta}(x, y) - \hat{s}_{HE}^{\theta}(x, y)}. \quad (14)$$

One might propose comparing our estimate $\hat{s}_{HE}^{\theta}(x, y)$ to a ground truth $s_{HE}^{\theta}(x, y)$ obtained using the pinhole-array technique. However, $\hat{s}_{HE}^{\theta}(x, y)$ is naturally estimated from data acquired at clinical exposure levels while $s_{HE}^{\theta}(x, y)$ must be acquired with non-clinical high exposure levels. The amount of scatter is exposure-dependent, so $\hat{s}_{HE}^{\theta}(x, y)$ and $s_{HE}^{\theta}(x, y)$ are incommensurate. However, the scatter-to-primary ratio $SPR_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{HE}^{\theta}(x, y)$ are exposure independent, and thus are commensurate and can be used for validation. Note that

the pinhole acquisitions are thus used for two purposes: to obtain the library data of section 2.5 and for validation.

Naturally, $SPR_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{HE}^{\theta}(x, y)$ should be derived from data on the same object. In particular, one would want to ideally compare $SPR_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{HE}^{\theta}(x, y)$ on patient data, but this is wholly impractical since this would entail multiple high-dose patient exposures with and without the pinhole plate mounted on the rotating scanner. Instead we compare $SPR_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{HE}^{\theta}(x, y)$ for the CIRS-020 phantom.

We performed validations, i.e. comparing $\hat{SPR}_{HE}^{\theta}(x, y)$ to $SPR_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{LE}^{\theta}(x, y)$ to $SPR_{LE}^{\theta}(x, y)$, using a CIRS-020 phantom with thicknesses 2,3,4,5 and 6 cm. A potential problem with using this phantom in that it has sharp edges, thus leading to possibly unrealistic penumbral and wing regions when compared to the rounded-edge of a real compressed breast. To address this, we performed another validation using the CIRS-011A phantom, a phantom with rounded edges designed to emulate a compressed breast (see Figure 2(c)). All validation results are reported in section 3.

For both phantom validation studies, in obtaining $\hat{SPR}_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{LE}^{\theta}(x, y)$, we used 66 mAs over 25 angles for HE acquisitions and 66-195 mAs for LE acquisitions, with the LE mAs varying with phantom thickness. For HE, we were limited to 66 mAs, a value approximately in a clinical range for a 4 cm thick breast. Using an exposure above 66 mAs on our Siemens Mammomat system resulted in a loss of signal in the wing regions. The signal loss is due not to detector saturation but to a “clipping” operation in which readings above a particular constant are set to the constant (The clipping operation was inherited from a previous version of our prototype detector and included for cosmetic display purposes. It will be eliminated in future prototype versions.). For both phantoms and both energies and for all angles, $SPR_{HE}^{\theta}(x, y)$ and $SPR_{LE}^{\theta}(x, y)$ were calculated at $3\times$ the dose used for $\hat{SPR}_{HE}^{\theta}(x, y)$ and $\hat{SPR}_{LE}^{\theta}(x, y)$. We repeated acquisitions three times and took the average.

2.7. Effects of scatter correction on reconstructed image quality

Our scatter correction method was tested on a CIRS-020 phantom with one slab containing iodine inserts. We acquired HE and LE images and performed an OS-SART reconstruction as described in section 2.2. A reconstructed slice was used to test the visibility of iodinated lesions with and without our scatter correction method. The 1 cm thick slab with inserts was always positioned is the second slab from the bottom.

We evaluated for the CIRS-020 phantom the degree of decoupling afforded by SC, using a metric (Siewerdsen *et al.*, 2006)

$$t_{cup}=100\frac{\bar{\mu}_{edge}-\bar{\mu}_{center}}{\bar{\mu}_{edge}} \quad (15)$$

where $\bar{\mu}_{edge}$ is a pixel average from five regions along the periphery of the reconstructed image and $\bar{\mu}_{center}$ a pixel average in a center region. Detailed results are given in section 3.3.

We report in section 4 anecdotal results of a CE-DE-DM patient image acquired under mammographic acquisition with a CC view. The images with and without our scatter correction are shown. The patient had a biopsy-confirmed tumor that was clearly visible in the image with SC but invisible without SC.

3. Results

In section 3.1 we validate our scatter correction algorithm on the CIRS-020 phantom by comparing $\hat{SPR}_{HE}^{\theta}(x, y)$ to $SPR_{HE}^{\theta}(x, y)$ (and $\hat{SPR}_{LE}^{\theta}(x, y)$ vs. $SPR_{LE}^{\theta}(x, y)$) at selected locations, for a variety of angles and a range of thicknesses. In section 3.2 this validation is repeated for the rounded edge CIRS011A phantom. In section 3.3 we show HE, LE projections and weighted subtraction reconstructions of the CIRS phantoms with and without SC. We report decoupling results using eq.(15). As explained in section 2.3.2, the choice of polynomial order for interpolating wing data is a crucial part of our SC algorithm. In section 3.1 we confirm that a fourth-order polynomial is optimal. We do this by comparing $\hat{SPR}_{HE}^{\theta}(x, y)$ vs. $SPR_{HE}^{\theta}(x, y)$ using the same methods as used in section 3.1, but with $\hat{SPR}_{HE}^{\theta}(x, y)$ assessed at a range of polynomial orders. We display reconstructions vs. polynomial order. Results from the clinical application of SC method are presented in section 3.3.

3.1. Validation results for CIRS-020 phantom

We report SPR and $\hat{SPR}$ results for HE and LE, for various thicknesses and for 3 representative angles: 0° , 12° and 23° . The values for SPR and $\hat{SPR}$ can be compared at all 225 pinhole locations, but we select a subset of 10 representative locations for each angle. The projection shadow is itself elongated at high θ relative to that at $\theta = 0^{\circ}$ and the plate orientation is θ -dependent. Both effects lead to a shift of the pinhole shadows as θ varies. The diagram in Figure 10 illustrates. Figure 10(a) shows a 0° acquisition illustrating the breast shadow region, pinhole locations (dots) and the 10 selected sampling points. Figure 10(b) shows the situation at $\theta = 23^{\circ}$ with the shadow now slightly elongated, the pinhole shadows slightly shifted relative to $\theta = 0^{\circ}$, and the 10 sampling points indicated.

The sampling points are numbered with indices 1-10, with points 1-5 corresponding to breast edge locations, 1,9,5 to locations along the chest wall, and 6,7,8 to interior locations. Note that while the positions of the 10 sampling points and the breast shadow itself both vary slightly with θ , the relative dispositions of the 10 index points vary only slightly.

Location #10 always corresponds to $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$.

Table 3 displays the HE (left) and LE (right) validation results for various thicknesses and angles. The first column in each sub-table indicates the sampling location via the id number as shown in Figure 10. The validations appear as adjacent columns labeled “ SPR ” and “ $\hat{SPR}$ ”. The subscripts (HE or LE), superscript (θ), and argument (x, y) associated with symbols SPR and $\hat{SPR}$ are accounted for by the HE and LE labels on the left and right tables,

the angle (0° , 12° or 23°) is shown above each SPR vs. $\hat{SPR}$ column pair, and the (x,y) location by the location id in the first column. The error bars on SPR and on $\hat{SPR}$ measurement pairs were far less (much less than 1%) than the difference between the SPR and $\hat{SPR}$ pair, so no standard deviations on SPR or $\hat{SPR}$ are listed. Note that in the 10th row of each table (corresponding to the point with id#10), SPR exactly equals $\hat{SPR}$. This is true by construction since as stated in the 2nd paragraph of section 3.1, location #10 always corresponds to $(\bar{x}_c^\theta, \bar{y}_c^\theta)$.

For HE at all thicknesses, the scatter estimates at 0° are quite accurate (with SPR and $\hat{SPR}$ differences 1% ~ 3%). As θ increases, accuracy drops (with differences up to 10%), especially for sampling points 4 and 5 corresponding to the region where the breast shadow is elongated. For LE comparison of SPR with $\hat{SPR}$, it shows good agreement except at points near the phantom edge. The discrepancy is due to the fact that our simple method of approximating $\hat{s}_{LE}^\theta(x, y)$ as a space-invariant constant leads to a discrepancy at edge points, where the true $s_{LE}^\theta(x, y)$ shows slow spatial variation (Sechopoulos *et al.*, 2007; Feng *et al.*, 2014; Boone and Cooper III, 2000).

Table 4 summarizes the results of Table 3 in the following way: it gives the mean relative error r_{HE}^- , defined as

$$\bar{r}_{HE} = \left\langle \left| \frac{\hat{SPR}_{HE}^\theta(x, y) - SPR_{HE}^\theta(x, y)}{SPR_{HE}^\theta(x, y)} \right| \right\rangle_{\theta, id} \quad (16)$$

A similar definition applies to r_{LE}^- . Both quantities are averaged over the angles and locations of Table 3 as indicated by the angle brackets.

The validations have used our SC algorithm of section 2.3.2 which entails a 4th-order polynomial interpolation of HE wing data. To validate that order 4 is optimal, we computed r_{HE}^- as a function of polynomial order and thickness. The results in Table 5 show that orders 4 and 5 are comparable for 2 cm and 4 cm thickness, but that order 4 is far superior (4.3% vs. 22.6%) for the thicker 6 cm phantom. We observed that for low orders (2 and 3), the error is due to underfitting - the quadratic or cubic interpolant does not have the shape flexibility to capture the scatter profile. For higher orders (6, 7 and 8) the error is due to overfitting. Order 5 also suffers overfitting for the thick 6 cm case. Note that the results of constant scatter subtraction for HE, discussed earlier, yields poor results as seen in the first column.

3.2. Validation with rounded edge phantom CIRS-011A

We also performed validations with the CIRS-011A phantom, described in section 2.1. It lacks iodine inserts, but its rounded edges emulate the shape of a compressed breast. Since the edge structure can affect the penumbral and wing region images used in our SC algorithm, the CIRS-011A serves as a good test. We repeated the validation steps of section 3.1 for the CIRS-011A (albeit only for the single thickness of 4.5 cm). We used a 10-point

sampling scheme as before with the points placement similar to that shown in Figure 10. The results, in a format that follows that of the tables in section 3.1, are presented in Table 6.

As with the CIRS-020 phantom, the accuracy of the SC algorithm is better for interior points than for edge points. Table 7 shows the mean relative errors r_{HE}^- and r_{LE}^- defined in eq.(16).

3.3 Projection and reconstruction results

The task of the SC is to improve the visibility of the iodine signal in the weighted subtraction reconstruction. In this section, we illustrate the efficacy of the SC algorithm with quantitative and qualitative results using the iodinated CIRS-020 phantom.

Figure 11 illustrates the scatter corrections for the iodinated CIRS-020 phantom by displaying projection images acquired under an assortment of conditions. For HE projection

images, we display $-\log \left[I_{HE}^\theta(x, y) / \tilde{b}_{HE}^\theta(x, y) \right]$ as the non-SC version and $-\log \left[\left(I_{HE}^\theta(x, y) - \hat{s}_{HE}^\theta(x, y) \right) / \tilde{b}_{HE}^\theta(x, y) \right]$ for the SC version. The LE versions follow similarly. The HE images suffer from a cupping effect that is mostly corrected by the scatter correction. Scatter correction improves the contrast of the LE images. Note that in the HE images of Figure 11, the non-SC and SC images (Figure 11(a) and (b)) clearly show decupping. However, inspection of the steep ($\theta = 23^\circ$) acquisitions of Figure 11(c) and (d) show that decupping uncovers a shading artifact (increasing intensity left to right) in the SC image. This shading is expected and is due to the radiometry of the steep-angle cone-beam acquisition.

Since scatter effects are most pronounced at high energy and the final reconstruction is a weighted subtraction of HE and LE reconstructions, we separately display HE reconstructions in Figure 12. The main effect of the SC is to eliminate the cupping artifacts along the breast periphery. As seen in Figure 12, as thickness increases, the cupping artifact becomes more severe.

Figure 13 shows a weighted subtraction in the reconstruction domain for the CIRS-020 phantom at 2, 4, and 6 cm. The weighted subtraction reconstruction is designed to suppress the anatomical variation, and comparison of the Figure 13 reconstructions to the HE-only reconstructions in Figure 12 shows that the “swirl” pattern due to anatomical variability is suppressed. The cupping artifact propagates into the reconstruction, but SC greatly reduces the degree of cupping. For the thicker 4 and 6 cm phantoms the decupping is greatly improved but incomplete and a rim of bright pixels remains.

The SC reconstructions in Figure 13 used a 4th order polynomial for interpolating the HE wing data. In Table 5 we showed that order 4 was optimal. Figure 14 displays weighted subtraction reconstructions of the 4 cm CIRS-020 phantom as a function of polynomial order and for the constant subtraction strategy.

Two important features of SC in the subtracted reconstruction are that: (1) The iodine signal intensity in a given slice becomes independent of the phantom thickness. (2) The decupping due to SC leads to improved iodine signal visibility in a given reconstructed slice.

The first feature is demonstrated in Figure 15 which plots iodine signal intensities in the slice containing iodine inserts reconstructed from phantoms of differing thickness. Signal intensities are computed by averaging pixels in signal regions as displayed in Figure 2(b). The pixel averages are taken only up to a radius 0.8 that of the full signal radii in order to avoid edge effects. Different thicknesses yield different amounts of scatter, which, if uncorrected, propagates into the subtracted reconstruction and alters signal values. Our SC does not yield absolute iodine quantitation in the subtracted reconstruction; this goal is intrinsically difficult due to the limited angle nature of DBT reconstruction (Puong *et al.*, 2008). However, by removing thickness dependence from iodine signal intensity, our SC may support improved iodine quantitation.

Figure 16 shows profiles of the reconstructions of the 4 cm thick phantom of Figure 13(b). The profiles clearly show the decoupling due to SC in the reconstructed image. In addition, the signal contrast is improved in the SC profiles.

We evaluated the degree of decoupling afforded by 4th-order polynomial SC, the constant SC strategy and no SC. To assess decoupling we used the metric t_{cup} of eq.(15). In eq.(15), μ_{center} is the pixel average over a circular region of radius 40 pixels centered on the midline, 4.5 cm from the chest wall. The average μ_{edge} is over 5 circular regions located along the phantom periphery and centered 0.85 cm from the edge. The 5 regions are equiangularly spaced at 45° , and each of the 5 circular regions is the same size used for μ_{Center} . None of the 6 regions overlap the positions of the iodine inserts. We averaged t_{cup} over 5 slices including the reconstructed slice in Figure 13 ± 2 slices, except for 2cm case where we evaluated 5 slices from the top since the top slice is the signal plane shown in Figure 13. For each of the averaged t_{cup} values, we computed a standard deviation using the individual t_{cup} values in each slice. Table 8 shows the results for phantoms of thicknesses 2, 4 and 6 cm. As seen in Table 8, the degree of 4th-order polynomial decoupling is nearly perfect for the 4 cm case, slightly under-decoupled for 6 cm and slightly over-decoupled for 2 cm. The use of the constant SC strategy yields results comparable to no SC.

In this section we have illustrated the decoupling effects of SC on projection data and reconstructions and we demonstrated that SC removes breast thickness dependence from the iodine signal intensity.

4. Anecdotal clinical results for scatter correction applied to CE-DE digital mammography

We had an opportunity to apply our SC algorithm to one clinical case for a patient previously diagnosed positive for breast cancer. Our protocol allowed only CE-DE mammography, not DBT. We performed a CE-DE mammographic CC view acquisition using 49 kVp for HE and 29 kVp for LE. We followed a patient-dose protocol that required use of automatic exposure control. The dose details are reported below. An HE/LE pair (0.3/1.15 mGy) was obtained at time 1 and a second HE/LE pair (0.21/0.61 mGy) at time 2 two minutes later. The contrast material might have changed slightly (diffused or concentrated) during this interval. The image pair at time 1 was obtained using an anti-

scatter grid. Importantly, at time 2 the HE image had no grid though the LE image was obtained with a grid. The patient's compressed breast thickness was 5.4 cm.

The fact that no grid was applied for the time 2 HE acquisition offered us an opportunity to test the SC algorithm. We performed a weighted subtraction of the time 2 HE acquisition (with our SC algorithm applied) and the time 2 LE acquisition with scatter correction afforded by a grid. The result is shown in Figure 17(c). Figure 17(b) shows the same result but with no SC algorithm applied to the HE acquisition. The anecdotal results are stunning, with the tumor clearly demonstrated in Figure 17(c) and absent in Figure 17(b). Reduction, due to the HE SC algorithm, of the cupping artifact in Figure 17(c) helps to reveal the tumor.

We can gain confidence that the results in Figure 17(b) and (c) are not artifactual by performing a weighted subtraction on the HE and LE time 1 acquisitions. Since these were obtained with a grid, they are scatter corrected. The result shows a lesion, along the left periphery, which is similar to that seen in Figure 17(c). The lesion in Figure 17(c) appears to be of higher contrast and slightly different structure compared to that in Figure 17(a). However, we do not know if this difference is due to contrast kinetics between time points 1 and 2.

5. Discussion and conclusion

We have developed an empirical SC algorithm for CE-DE-DBT. The results are encouraging but further work is needed to address some limitations of our study. One important limitation - application of SC to the MLO view - is discussed at length below in section 5.1. Other limitations are discussed in section 5.2.

5.1 Application of SC to MLO view

Our studies entailed only CC views and our SC algorithm must be adapted for MLO views. The geometry of the MLO projection view eliminates some of the wing regions. To see this, examine the clinical LE MLO mammogram in Figure 18 (a). For the CC view, we used one-dimensional polynomial interpolation along the x direction parallel to the chest wall. In Figure 18 (a) the chest direction (x-direction) runs vertically as shown. Thus, as seen in Figure 18 (a), if we interpolate along x, there is a wing region on both sides of the breast edge for only a fraction of the breast. For much of the breast, the wing region is available only on one side.

Below, we show results of a preliminary study that uses a strategy to circumvent the “one-side wing” problem. We used the clinical image in Figure (a) to design a crude 4 cm thick MLO phantom for which we acquired HE projection data (using the same mAs and kVp as in the CC case) at $\theta = 0^\circ$ and at $\theta = 23^\circ$. The $\theta = 23^\circ$ view is the worst possible MLO view in our $[-23^\circ, +23^\circ]$ range since it leads to the maximum exclusion of interpolatable wing regions. We apply our modified SC algorithm to these two HE views. Recall that no modification is needed for LE data since wing interpolation is not used.

The MLO phantom, shown at 0° in Figure 18 (b), comprises a translated and rotated 4 cm CIRS-020 semi-circular phantom that abuts a 4 cm thick Lucite square. The translation and

rotation of the CIRS-020 is designed to emulate the breast truncation in the MLO view, and the Lucite phantom, only a corner of which appears, is designed to emulate the pectoral muscle. The pectoral muscle has greater attenuation than breast, and the Lucite has correspondingly greater attenuation than the CIRS-020 phantom. The geometrical arrangement of the phantom in Figure 18 (b) is designed to follow the structure of the clinical phantom in Figure 18 (a) and corresponding characteristic geometrical points 1, 2, 3 and 4 are shown in each image. In particular, the line connecting points 1 and 3 delineates the pectoral muscle and the line connecting points 1 and 2 shows how the MLO view is truncated by the Tail of Spence - tissue that forms a ridge pointing toward the upper outer breast quadrant. The line connecting points 3 and 4 shows a typical boundary beyond which the lowest part of the breast can be missed in the MLO view.

The MLO SC algorithm is explained with the aid of Figures 18 (c)(d). We begin by determining points A and B (corresponding to points 2 and 4 in the clinical image). If one translates line $\overline{AB}$ perpendicular to itself by an offset, shown by the double arrow, then a parallel line $\overline{A'B'}$ is formed. The offset is chosen to allow enough space to fit wing regions shown as the dotted lines beyond A' and B' . For the wing regions, a distance of 50 pixels is enough for effective fourth-order polynomial interpolation. Polynomial interpolation is then possible along all lines parallel to $\overline{A'B'}$ lying on the nipple side. A few such lines are shown (in purple) in Figure 18 (c) and the locus of all such lines is indicated by the purple region in Figure 18 (d).

However, lines parallel to $\overline{A'B'}$ on the chest wall side (indicated by blue lines in Figure 18 (d)) are unavailable for interpolation since one or both wings are not present. As an approximation, we replicate the $\overline{A'B'}$ interpolation on the set of (blue) lines in this “forbidden” region. The justification for this approximation is as follows: Due to truncation near the chest wall in the MLO view, lines in the forbidden region are mostly far from the breast edge, where cupping is prominent. Along the set of lines in the forbidden region far from the breast edge, the interpolated scatter estimates vary slowly. Therefore replication of the interpolant along the $\overline{A'B'}$ line may not be too inaccurate. Given all the 1-D polynomial interpolations, we smooth in a direction perpendicular to $\overline{A'B'}$ and also add the offset k as we did for the CC view.

Figures 18 (e)-(h) show the uncorrected and scatter corrected projection data at $\theta = 0^\circ$ and at $\theta = 23^\circ$. It is seen that the cupping artifact in this HE projection data, which is the main problem caused by scatter, is greatly diminished. We have shown an approach to MLO SC with good anecdotal results. Computation of critical points A , B , A' , B' and offset are easy to do. Further work in validating this approach is needed.

5.2 Other limitations

Since the SC algorithm depends on interpolation of wing data in the HE projection images, and since the wing data lies in a high signal region outside the breast image, it is important that the region outside the breast image not be distorted due to detector saturation, detector nonlinearity, or other effects such as “clipping” operation discussed in section 2.6. Any distortions of the wing data can yield large errors in the polynomial interpolation which will result, in turn, an inaccurate scatter estimates.

The SC algorithm needs further improvement in estimating scatter in the penumbral region. Inspection of the SC reconstructions in Figure 13 (right column) shows a thin bright ring artifact along the breast periphery that is worse for thicker breasts. The SC algorithm overestimates scatter in the penumbral region as seen in Figure 4(d), and this overestimate leads to the artifact.

One might suspect that SC could improve absolute or relative quantitation of iodine concentration in the reconstruction. However, Puong et. al. (Puong *et al.*, 2008) show that iodine quantitation is intrinsically difficult for CE-DE-DBT. The difficulty is due to the severely limited angular acquisition of DBT. However, our results in Figure 15 show one useful aspect of SC: the reconstructed iodine voxel values become independent of breast thickness.

In CE-DE-DBT, the visualization of iodinated lesions is improved by anatomical background suppression and K-edge enhancement of the iodine signal. Nevertheless, cupping effects may lead to masking effects for iodine signals located along the breast periphery. Scatter correction can be quite useful in addressing this problem.

Finally, one might avoid the complexities of algorithmic SC by using an anti-scatter grid. For tomosynthesis, the grid vanes would need to be oriented in the y-direction to avoid angular blocking. However, any such use of grids entails a dose penalty due to absorption of primary photons by the grid.

5.3 Conclusion

We presented an algorithm designed to accomplish rapid patient-specific scatter correction for CE-DE-DBT. For a given site, it requires simple calibration measurements obtained from a pinhole SPR measurement of a standardized phantom. We validated its accuracy for breast emulating phantoms. The resulting scatter estimate was shown to reduce cupping artifacts in reconstructions and to remove the effect of phantom thickness on reconstructed values of iodine inserts.

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Figure 1.
Siemens Mammomat Inspiration CEDET scanner used for acquisitions. A CIRS BR3D (model 020) phantom is shown mounted in the scanner. The inset shows a customized slab of the phantom with iodine inserts.

Figure 2. CIRS-020 and CIRS-011A phantoms

(a) Photo of 9 cm radius slab with iodine inserts. (b) Drawing to scale of signal region of CIRS-020 shows the iodine insert indices. (c) The CIRS-011A phantom comprises a skin tissue cover (as shown in white) and breast mimicking tissue component (as shown in pink).

Figure 3. Stylized DBT system (not to scale)

The x-y-z coordinates are defined in (a). The x-y plane is coincident with the detector plane. A breast sits atop the detector and is irradiated by an X-ray tube that rotates by θ in the x-z plane. The center-of-rotation sits above the x-y plane as shown in (b). Configurations at zero degrees (a) and at an oblique angle (b) are shown. Profiles in the x-direction at a fixed y indicate the detector readout in three regions: object, penumbra and wing, defined in the text.

Figure 4. Illustration of 1-D version of scatter estimation

(See text for detailed explanation). (a) True scatter profile. (b) Green curve is polynomial interpolation from wing data. (c) In object region, interpolated profile differs from true scatter by additive offset k . (d) Blue curve, obtained by adding k to the interpolated green curve, is the scatter estimate. It is accurate in the object region and an inaccurate in the penumbral region.

Figure 5. Obtaining initial scatter estimate from wing data

In (a), the segmented and smoothed wing data (cyan) and its 4th-order polynomial interpolation (green) are shown for one value of y and for $\theta = 0^\circ$. In (b), the interpolated curves $\hat{s}_{HE}^\theta(x, y)$ for all y are shown as a greyscale image. The green line corresponds to the 1-D profile of scatter in (a). Further smoothing in the y -direction results in the image $\hat{s}_{HE}^\theta(x, y)$ in (c).

Figure 6. Profiles in the x-direction of the various stages of the SC algorithm applied for HE acquisition of a CIRS-020 phantom

Results are shown for two angles. The algorithm takes the raw data I_{HE}^{θ} (in blue) and ends with the scatter corrected data p_{HE}^{θ} (in red).

Figure 7. Pinhole array acquisition

- (a) Diagram (not to scale) shows acquisition with plate oriented at θ . Only the plate and phantom are shown. The counts in the pinhole shadow will comprise primary-only photons.
- (b) The lead plate with a 15×15 array of holes.

Figure 8.

Partial profile of image of a Lucite phantom HE acquisition with (a) plate inserted (red), showing the image due to three adjacent pinholes and (b) with plate removed (blue). For a given hole region, “Primary” indicates the contribution due to primary photons and “Scatter” due to scattered photons.

Figure 9. Test of pinhole shadow interactions

The red curve is a profile of 3 of the 225 simultaneously irradiated pinholes and the blue curve the profile obtained from irradiating the reference hole only. The left and right peaks are from pinholes that neighbor the reference pinhole. See text for a quantitative analysis.

Figure 10. Sampling locations for validation

The breast shadow region (solid line) and pinhole shadow (dots) are drawn to scale for a 0° acquisition in (a) and a 23° acquisition in (b). A subset of 10 pinhole locations, indicated by circles, is chosen at each angle and validation is done at these locations. Each of the 10 pinhole locations is labeled by an index number.

Figure 11. Projection images with and without scatter correction for 4cm thick CIRS-020 phantom

Top row: HE images. Bottom row: LE images. Images (a) and (e) at $\theta = 0^\circ$ are non-scatter-corrected, (b) and (f) corrected at $\theta = 0^\circ$, (c) and (g) non-corrected at $\theta = 23^\circ$, and (d) and (h) corrected at $\theta = 23^\circ$. Each SC and no-SC pair uses the same grey scale display window width though the windows are shifted to enhance visual clarity.

Figure 12. Reconstruction of HE data with and without SC

Shown are the reconstructed slices of the CIRS020 phantom at (a) 2cm (b) 4cm and (c) 6cm. The left, right images of each pair are no SC and SC, respectively.

Figure 13. Weighted subtraction in the reconstructed domain for CIRS-020 phantom
We use HE and LE reconstruction followed by weighted subtraction to derive the reconstructed slices with iodine inserts. (a), (b) and (c) are for the CIRS-020 phantom at 2, 4 and 6 cm, respectively. The left image of each pair is for no SC and the right image uses SC data.

Figure 14. Weighted subtraction SC reconstructions using different polynomials orders
The images above are all reconstruction-then-subtraction results for the 4cm CIRS-020 phantom. From left to right and top to bottom, the orders of polynomial used to do wing interpolations are 0, 2, 3, 4, 5, 6, 7 and 8, respectively. Order “0” means we use the constant scatter subtraction strategy for HE and LE data. The weighting factor and grey-scale windows are identical.

Figure 15. Reconstructed intensities of iodine inserts

The ordinate is the reconstructed signal value and the abscissa the index of the signal as displayed in Figure 2(b). (a) No SC. Signals 1, 5, 9, 13 for thickness 2, 4, 6 cm. (b) Same as (a) but with SC. (c) No SC. Signals 4, 8, 12, 16 for thicknesses 2, 4, and 6 cm. (d) Same as (c) but with SC. Note that in (b) and (d), the curves are coincident, illustrating the independence of signal with thickness.

Figure 16. Profiles of subtracted reconstruction in signal slice for 4 cm phantom

(a) Profile along centers of top row iodine inserts. (b) Profile along centers of bottom row of iodine inserts.

Figure 17. Patient image for CE-DE mammography

(a) Weighted subtraction at time 1 with SC for HE and LE provided by the grid (b) Weighted subtraction at time 2 projection data with no SC applied to the HE data and grid SC for the LE data (c) Same as (b) but with our SC algorithm applied to the HE data.

Figure 18. Scatter correction for MLO view

(a) Typical clinical LE MLO mammogram (b) LE Mammogram of our MLO phantom designed to emulate that in (a). Points 1, 2, 3, and 4 in each image correspond. (c)(d) Trajectories for interpolation. See text for details. (e)(f) HE projection view at 0° with no SC (e) and with SC (f). (g)(h) HE projection at $+23^\circ$ with no SC (g) and with SC (h). Note the decoupling associated with SC.

Table 1**Symbol definitions**

The subscripts HE and LE refer to high energy and low energy. The superscript θ refers to the angle of acquisition. The coordinate (x,y) is position in the detector plane.

Symbols	Descriptions
$I_{HE}^{\theta}(x, y), I_{LE}^{\theta}(x, y)$	Raw projection data
$g_{HE}^{\theta}(x, y), g_{LE}^{\theta}(x, y)$	Projection data after log step
$b_{HE}^{\theta}(x, y), b_{LE}^{\theta}(x, y)$	Blank scan data
$\tilde{b}_{HE}^{\theta}(x, y), \tilde{b}_{LE}^{\theta}(x, y)$	Smoothed versions of $b_{HE}^{\theta}(x, y), b_{LE}^{\theta}(x, y)$
$w_{HE}^{\theta}(x, y)$	Smoothed wing data of HE acquisition
$s_{HE}^{\theta}(x, y), s_{LE}^{\theta}(x, y)$	True scatter
$\tilde{s}_{HE}^{\theta}(x, y)$	Interpolation of wing data along x for a fixed y (repeated for each y)
$\tilde{s}_{HE}^{\theta}(x, y), \tilde{s}_{LE}^{\theta}(x, y)$	Smoothed (in y direction) version of $\tilde{s}_{HE}^{\theta}(x, y)$. Note $\tilde{s}_{LE}^{\theta}(x, y)$ is set to zero
$k_{HE}^{\theta}, k_{LE}^{\theta}$	Constant used to correct $\tilde{s}_{HE}^{\theta}(x, y), \tilde{s}_{LE}^{\theta}(x, y)$
$\hat{s}_{HE}^{\theta}(x, y), \hat{s}_{LE}^{\theta}(x, y)$	Final estimated scatter
$SPR_{HE}^{\theta}(x, y), SPR_{LE}^{\theta}(x, y)$	True scatter-to-primary ratios
$\hat{SPR}_{HE}^{\theta}(x, y), \hat{SPR}_{LE}^{\theta}(x, y)$	Estimates of scatter-to-primary ratio derived using our scatter correction algorithm
$\overline{SPR}_{HE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta}), \overline{SPR}_{LE}^{\theta}(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$	Reference library <i>SPR</i> point value obtained from a phantom at location $(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$
$(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$	Reference point near center of breast, location where $\tilde{s}_{HE}^{\theta}(x, y)$ peaks
$(\bar{x}_c^{\theta}, \bar{y}_c^{\theta})$	Reference point near center of phantom, location where $s_{HE}^{\theta}(x, y)$ peaks
$p_{HE}^{\theta}(x, y), p_{LE}^{\theta}(x, y)$	Primary photon readings

Table 2
Pseudocode for HE SC Algorithm

for $\theta = \theta_{initial}$ to θ_{final} in steps of θ

$$\tilde{b}_{HE}^{\theta}(x, y) = b_{HE}^{\theta}(x, y) \otimes G_{15 \times 15}^{\sigma=7}(x, y)$$

$$w_{HE}^{\theta}(x, y) = \left[(I_{HE}^{\theta}(x, y) - \tilde{b}_{HE}^{\theta}(x, y)) \otimes G_{5 \times 5}^{\sigma=3}(x, y) \right]_{+} \text{ (} []_{+} \text{ means clip negative values to 0)}$$

for $y_{ind} = 1$ to (# of detector bins in y)

1-D 4th order polynomial interpolation through $w_{HE}^{\theta}(x, y_{ind})$ to compute $\tilde{s}_{HE}^{\theta}(x, y_{ind})$

end

for $x_{ind} = 1$ to (# of detector bins in x)

$$\tilde{s}_{HE}^{\theta}(x_{ind}, y) = \tilde{s}_{HE}^{\theta}(x_{ind}, y) \otimes G_{1 \times 200}^{\sigma=60}(y)$$

end

$$(x_c^{\theta}, y_c^{\theta}) = \arg \max (\tilde{s}_{HE}^{\theta}(x, y))$$

Look up $\overline{SPR}_{HE}^{\theta}(\overline{x}_c^{\theta}, \overline{y}_c^{\theta})$ from library

$$\hat{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) = I_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) \cdot \frac{\overline{SPR}_{HE}^{\theta}(\overline{x}_c^{\theta}, \overline{y}_c^{\theta})}{1 + \overline{SPR}_{HE}^{\theta}(\overline{x}_c^{\theta}, \overline{y}_c^{\theta})}$$

$$k_{HE}^{\theta} = \hat{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta}) - \tilde{s}_{HE}^{\theta}(x_c^{\theta}, y_c^{\theta})$$

$$\hat{s}_{HE}^{\theta}(x, y) = \tilde{s}_{HE}^{\theta}(x, y) + k_{HE}^{\theta} \text{ is the final scatter estimate}$$

end

Table 3
Validation results for scatter correction algorithm applied to CIRS-020 phantom

HE										LE									
0 degrees					12 degrees					0 degrees					12 degrees				
id	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR̂	id	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR̂
2 cm	1	23.60%	24.00%	29.30%	27.50%	29.90%	28.70%	28.70%	28.70%	1	25.40%	28.60%	28.70%	26.90%	33.40%	33.10%	33.40%	33.10%	33.10%
	2	29.60%	28.50%	31.00%	29.00%	32.50%	29.10%	29.10%	29.10%	2	31.90%	32.30%	33.70%	32.70%	34.40%	34.70%	34.40%	34.70%	34.70%
	3	30.50%	27.80%	29.20%	28.50%	29.50%	27.40%	27.40%	27.40%	3	28.30%	31.10%	31.00%	32.10%	38.10%	39.60%	38.10%	39.60%	39.60%
	4	29.40%	27.80%	33.80%	32.70%	33.80%	32.90%	32.90%	32.90%	4	28.30%	28.80%	32.70%	35.20%	33.40%	38.50%	33.40%	38.50%	38.50%
	5	25.10%	25.70%	31.10%	33.30%	28.10%	28.00%	28.00%	28.00%	5	24.80%	26.20%	31.90%	34.80%	33.30%	43.00%	33.30%	43.00%	43.00%
	6	34.00%	34.30%	37.50%	38.30%	38.60%	37.60%	37.60%	37.60%	6	31.60%	29.90%	31.40%	29.30%	31.60%	31.40%	31.60%	31.40%	31.40%
	7	36.30%	32.90%	37.00%	34.90%	38.20%	36.60%	36.60%	36.60%	7	30.90%	30.10%	30.30%	29.60%	37.30%	38.50%	37.30%	38.50%	38.50%
	8	35.00%	34.90%	39.50%	39.60%	39.60%	40.40%	40.40%	40.40%	8	31.00%	31.10%	36.20%	34.90%	37.80%	38.70%	37.80%	38.70%	38.70%
	9	35.40%	37.60%	38.40%	41.60%	36.80%	40.80%	40.80%	40.80%	9	28.50%	30.50%	32.50%	33.50%	29.80%	32.30%	29.80%	32.30%	32.30%
	10	37.10%	37.10%	39.50%	39.50%	43.70%	43.70%	43.70%	43.70%	10	32.40%	32.40%	34.50%	34.50%	35.80%	35.80%	35.80%	35.80%	35.80%
3 cm	1	35.80%	36.00%	41.60%	39.00%	44.40%	43.30%	43.30%	43.30%	1	40.60%	43.20%	42.30%	41.00%	45.10%	45.20%	45.10%	45.20%	45.20%
	2	40.20%	39.00%	43.40%	39.50%	43.80%	39.40%	39.40%	39.40%	2	45.60%	45.80%	38.90%	38.40%	43.50%	43.40%	43.50%	43.40%	43.40%
	3	36.80%	35.70%	38.70%	37.30%	39.70%	36.40%	36.40%	36.40%	3	41.00%	44.10%	46.60%	49.70%	51.40%	56.80%	51.40%	56.80%	56.80%
	4	36.60%	36.30%	42.80%	42.20%	45.60%	44.20%	44.20%	44.20%	4	43.90%	44.40%	45.70%	50.20%	52.40%	61.60%	52.40%	61.60%	61.60%
	5	33.00%	34.30%	38.50%	39.70%	38.00%	40.90%	40.90%	40.90%	5	32.30%	33.60%	39.60%	45.80%	49.00%	59.20%	49.00%	59.20%	59.20%
	6	46.40%	46.30%	51.40%	49.20%	48.90%	49.40%	49.40%	49.40%	6	40.70%	40.60%	40.20%	35.50%	52.50%	49.10%	52.50%	49.10%	49.10%
	7	48.10%	45.50%	48.00%	45.60%	49.40%	46.50%	46.50%	46.50%	7	39.40%	40.30%	44.30%	41.20%	42.90%	45.00%	42.90%	45.00%	45.00%
	8	46.40%	45.60%	51.80%	51.70%	51.50%	53.50%	53.50%	53.50%	8	39.70%	39.70%	43.50%	45.00%	55.50%	55.10%	55.50%	55.10%	55.10%
	9	45.00%	48.10%	49.60%	53.50%	45.00%	51.60%	51.60%	51.60%	9	44.40%	44.40%	48.00%	48.40%	37.90%	38.20%	37.90%	38.20%	38.20%
	10	50.30%	50.30%	54.40%	54.40%	54.00%	54.00%	54.00%	54.00%	10	43.70%	43.70%	41.40%	41.40%	53.10%	53.10%	53.10%	53.10%	53.10%
0 degrees					12 degrees					0 degrees					12 degrees				
id	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR̂	id	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR	SPR̂	SPR̂

LE											
HE						LE					
id	0 degrees		12 degrees		23 degrees	id	0 degrees		12 degrees		23 degrees
	SPR	SPR̂	SPR	SPR̂	SPR		SPR	SPR̂	SPR	SPR̂	SPR
1	46.40%	47.90%	46.20%	49.10%	50.90%	1	44.90%	48.70%	50.60%	48.00%	42.30%
2	48.30%	48.90%	53.30%	54.40%	55.20%	2	41.20%	41.30%	51.60%	48.40%	44.50%
3	53.20%	53.10%	52.00%	46.80%	51.80%	3	47.40%	49.10%	55.00%	61.70%	52.90%
4	48.90%	50.40%	46.80%	46.70%	56.40%	4	43.80%	43.20%	53.50%	57.60%	47.40%
5	45.40%	46.00%	48.50%	49.70%	45.10%	5	46.60%	48.60%	63.20%	75.10%	59.50%
6	57.40%	56.00%	61.20%	64.10%	61.60%	6	49.40%	48.60%	51.80%	51.90%	46.60%
7	59.40%	56.90%	65.10%	59.30%	64.60%	7	48.00%	48.90%	62.60%	61.80%	55.60%
8	59.00%	57.10%	59.00%	60.10%	67.70%	8	50.10%	47.70%	54.30%	52.90%	55.70%
9	59.50%	60.80%	64.20%	69.00%	65.40%	9	55.00%	51.60%	57.40%	60.00%	55.70%
10	62.40%	62.40%	66.20%	66.20%	65.40%	10	41.00%	41.00%	49.40%	49.40%	52.30%
4 cm											
id	0 degrees		12 degrees		22 degrees	id	0 degrees		12 degrees		23 degrees
	SPR	SPR̂	SPR	SPR̂	SPR		SPR	SPR̂	SPR	SPR̂	SPR
1	58.60%	65.20%	58.70%	58.60%	55.40%	1	64.20%	62.60%	65.60%	62.10%	73.90%
2	65.70%	68.00%	60.70%	58.20%	63.80%	2	55.90%	52.20%	66.60%	63.80%	56.50%
3	67.30%	68.70%	56.80%	57.50%	56.70%	3	74.60%	73.60%	77.70%	76.00%	79.30%
4	65.20%	68.30%	60.90%	63.00%	56.20%	4	46.20%	52.60%	52.90%	63.30%	69.80%
5	57.30%	63.00%	59.10%	65.70%	64.70%	5	49.40%	53.90%	76.00%	86.30%	76.90%
6	70.10%	71.00%	71.80%	68.30%	77.10%	6	60.90%	58.60%	57.20%	57.20%	72.40%
7	75.60%	73.90%	73.60%	69.30%	73.10%	7	53.10%	57.80%	66.60%	72.40%	84.40%
8	73.40%	73.30%	74.70%	74.30%	78.60%	8	59.70%	58.70%	62.10%	64.60%	77.30%
9	69.80%	75.20%	77.00%	80.70%	80.10%	9	57.60%	60.50%	61.80%	64.30%	75.20%
10	77.80%	77.80%	76.70%	76.70%	83.50%	10	58.90%	58.90%	59.30%	59.30%	73.50%
5 cm											
id	0 degrees		12 degrees		23 degrees	id	0 degrees		12 degrees		23 degrees
	SPR	SPR̂	SPR	SPR̂	SPR		SPR	SPR̂	SPR	SPR̂	SPR
1	62.70%	67.30%	72.90%	74.00%	83.10%	1	75.50%	79.40%	71.10%	67.40%	77.10%
2	66.90%	71.00%	72.30%	71.80%	67.40%	2	85.00%	83.10%	83.40%	72.50%	69.60%
6 cm											
1	62.70%	67.30%	72.90%	74.00%	83.10%	1	75.50%	79.40%	71.10%	67.40%	77.10%
2	66.90%	71.00%	72.30%	71.80%	67.40%	2	85.00%	83.10%	83.40%	72.50%	69.60%

LE														
0 degrees					12 degrees					23 degrees				
id	SPR	SPR̂	SPR	SPR̂	id	SPR	SPR̂	SPR	SPR̂	id	SPR	SPR̂	SPR	SPR̂
3	73.50%	76.30%	68.60%	70.60%	3	81.60%	79.00%	82.80%	80.20%	3	81.60%	79.00%	106.00%	99.00%
4	76.00%	75.50%	73.70%	78.20%	4	74.50%	71.70%	80.30%	90.30%	4	74.50%	71.70%	99.70%	127.90%
5	68.40%	71.20%	71.20%	79.00%	5	72.30%	77.50%	77.20%	104.50%	5	72.30%	77.50%	91.90%	128.60%
6	81.70%	80.00%	81.00%	81.80%	6	67.60%	67.20%	92.10%	85.40%	6	67.60%	67.20%	91.80%	89.20%
7	81.70%	79.50%	81.60%	78.90%	7	74.10%	68.60%	70.50%	70.90%	7	74.10%	68.60%	91.10%	91.50%
8	81.20%	79.90%	89.50%	89.20%	8	79.00%	80.30%	73.30%	75.90%	8	79.00%	80.30%	94.50%	111.70%
9	80.70%	81.50%	83.70%	91.70%	9	75.00%	76.70%	79.20%	82.50%	9	75.00%	76.70%	86.20%	94.30%
10	83.30%	83.30%	88.80%	88.80%	10	68.70%	68.70%	79.00%	79.00%	10	68.70%	68.70%	80.80%	80.80%

Table 4
Relative averaged error of SPR and $\hat{SPR}$ for CIRS-020

	r_{HE}	r_{LE}
2cm	4.16%	5.21%
3cm	4.11%	4.79%
4cm	4.37%	4.93%
5cm	4.70%	5.98%
6cm	4.25%	8.18%

Table 5

Relative averaged error as a function of polynomial order

Entries are r_{HE} for various polynomial orders and the constant subtraction strategy and for three thicknesses of the CIRS-020 phantom. An entry of X corresponds to scatter interpolants unstable enough to produce negative scatter estimates.

	Constant	Order 2	Order 3	Order 4	Order 5	Order 6	Order 7	Order 8
2 cm	18.93%	13.98%	13.95%	4.16%	3.57%	26.18%	21.73%	38.71%
4 cm	19.10%	11.98%	12.39%	4.37%	4.83%	8.42%	18.85%	48.02%
6 cm	17.01%	9.40%	11.30%	4.25%	22.62%	31.07%	X	X

Table 6
Validation results for scatter correction algorithm applied to CIRS-011A phantom

LE									
HE									
id	0 degrees			12 degrees			23 degrees		
	SPR	SP $\hat{R}$	R	SPR	SP $\hat{R}$	R	SPR	SP $\hat{R}$	R
1	46.5%	51.3%	54.6%	55.1%	55.9%	55.2%	55.5%	51.9%	53.5%
2	53.5%	52.8%	56.6%	55.0%	55.9%	50.8%	54.5%	54.5%	58.5%
3	43.8%	41.6%	64.8%	64.7%	67.3%	68.6%	55.0%	35.6%	37.5%
4	52.8%	51.5%	64.9%	71.4%	69.5%	81.0%	55.0%	46.5%	48.1%
5	49.5%	53.2%	60.4%	75.2%	67.0%	86.8%	53.4%	46.4%	48.8%
6	54.2%	53.4%	63.0%	60.6%	63.5%	61.4%	65.3%	70.7%	69.8%
7	54.1%	51.8%	65.1%	66.0%	73.1%	72.9%	62.8%	61.7%	67.8%
8	51.1%	52.4%	69.4%	70.8%	72.7%	75.2%	69.2%	73.5%	73.2%
9	52.5%	53.4%	58.6%	63.7%	65.9%	68.5%	67.8%	71.2%	71.0%
10	52.7%	52.7%	64.0%	64.0%	69.7%	69.7%	69.9%	75.2%	75.2%

Table 7
Relative averaged error of SPR and $SP\hat{R}$ for CIRS-011A

	$\bar{r}_{HE}$	$\bar{r}_{LE}$
4.5cm	4.02%	5.34%

Table 8
Decoupling results, t_{cup} for CIRS020 reconstructed images

	2cm	4cm	6cm
No SC	$23.03\% \pm 2.22\%$	$24.42\% \pm 2.89\%$	$18.81\% \pm 2.59\%$
Constant SC	$23.96\% + 2.42\%$	$27.41\% + 2.61\%$	$21.44\% + 2.51\%$
SC	$-7.52\% \pm 1.98\%$	$-2.19\% \pm 1.79\%$	$9.33\% \pm 3.01\%$