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Large-angle x-ray scatter in Talbot-Lau interferometry for breast imaging

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Abstract

Monte Carlo simulations were used to investigate large-angle x-ray scatter at design energy of 25 keV during small field of view (9.6 cm × 5 cm) differential phase contrast imaging of the breast using Talbot-Lau interferometry. Homogenous, adipose and fibroglandular breasts of uniform thickness ranging from 2 to 8 cm encompassing the field of view were modeled. Theoretically determined transmission efficiencies of the gratings were used to validate the Monte Carlo simulations, followed by simulations to determine the x-ray scatter reaching the detector. The recorded x-ray scatter was classified into x-ray photons that underwent at least one Compton interaction (incoherent scatter) and Rayleigh interaction alone (coherent scatter) for further analysis. Monte Carlo based estimates of transmission efficiencies showed good correspondence ($r^2 > 0.99$) with theoretical estimates. Scatter-to-primary ratio increased with increasing breast thickness, ranging from 0.11 to 0.22 for 2 to 8 cm thick adipose breasts and from 0.12 to 0.28 for 2 to 8 cm thick fibroglandular breasts. The analyzer grating reduced incoherent scatter by ~18% for 2 cm thick adipose breast and by ~35% for 8 cm thick fibroglandular breast. Coherent scatter was the dominant contributor to the total scatter. Coherent-to-incoherent scatter ratio ranged from 2.2 to 3.1 for 2 to 8 cm thick adipose breasts and from 2.7 to 3.4 for 2 to 8 cm thick fibroglandular breasts.

Keywords

Breast; Mammography; X-ray phase contrast; X-ray scatter; Interferometry

I. Introduction

X-ray phase contrast imaging of the breast using Talbot-Lau interferometry that uses a conventional x-ray source and three gratings is being actively investigated by several research groups (Pfeiffer et al., 2006, Stampanoni et al., 2011, Zysk et al., 2012, Bevins et al., 2012, Tang et al., 2012). The three gratings are the source grating (G0) to provide multiple coherent beams, the phase or beam-splitter grating (G1) that produces Talbot self-images at periodic distances from the grating, and the analyzer or absorption grating (G2)

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that facilitates recording of the fringes using practical x-ray imaging detectors by demodulating the fringe function from the fringe displacement (Pfeiffer et al., 2006, Fredenberg et al., 2012). When x-rays traverse through an object with refractive index $n = 1 - \delta + i\beta$, they undergo refraction and attenuation that is dependent on the refractive index decrement δ and attenuation index β , respectively. Talbot-Lau interferometry provides the transmission or attenuation image $T(x, y)$ that is dependent on β and is similar to a conventional mammogram, the normalized fringe visibility or the dark-field image $V(x, y)$ arising from small angle x-ray scattering of particulates or microstructures of scale much smaller than the resolution of the system (Pfeiffer et al., 2008, Nesterets, 2008, Bech et al., 2010, Yashiro et al., 2010, Chen et al., 2010, Lynch et al., 2011, Yang and Tang, 2012, Michel et al., 2013, Willner et al., 2014, Wang et al., 2014), and the differential phase

contrast image $\frac{\partial \Phi(x, y)}{\partial x}$ or $\Phi'(x, y)$ that is related to the measured phase shift $\phi_1(x, y)$ and can be represented as (Chen et al., 2011):

$$\phi_1(x, y) = \frac{\lambda d}{p_2} \frac{\partial \Phi}{\partial x} = - \frac{2\pi d}{p_2} \frac{\partial}{\partial x} \int \delta(x, y, z) dz \quad (1)$$

In the above equation, λ is the wavelength of x-rays, d is the spacing between the phase (G1) and analyzer (G2) gratings (Fig 1), p_2 is the pitch of the G2 grating, and the x-ray beam is propagated along the z -direction. If a uniformly thick, homogenous breast encompassing the FOV of the interferometer is considered, then $\delta(x, y, z) = \delta$, $\Phi'(x, y) = 0$ from equation (1). Also, $V(x, y) = V$ (Lynch et al., 2011). Under these conditions, a Monte Carlo toolkit

modeling the linear attenuation coefficient $\mu = \frac{4\pi\beta}{\lambda}$ can be used to determine x-ray scatter without the confounding effects of scatter arising from tissue structure.

At 25 keV and using elemental composition for adipose and fibroglandular tissue (Hammerstein et al., 1979), the percent contribution to the total mass attenuation coefficient (Berger et al., 2010) from photoelectric interaction, incoherent scattering and coherent scattering are 38%, 49%, and 13%, respectively for adipose tissue, and 50%, 38%, and 12% for fibroglandular tissue (Vedantham and Karellas, 2013). Hence, the probability of an x-ray photon undergoing scattering, in particular incoherent scattering, is substantial. This study was conducted to determine as to whether: (1) the coherent scatter was the dominant contributor to the total x-ray scatter recorded by the detector; and, (2) the G2 grating reduces the number of scattered x-ray photons, in particular incoherent scattered photons, reaching the detector. Additionally, the intent was to provide an estimate of scatter-to-primary ratio incident on the detector, so that its effect can be modeled during numerical simulations. Hence, Monte Carlo (MC) simulations that allow inclusion of multiple scattering were conducted to estimate x-ray scatter arising from the components of the imaging setup for a realistic cone-beam imaging geometry, and after introduction of 2 to 8 cm thick homogenous adipose and fibroglandular breasts in the beam path.

II. Methods and Materials

Initial studies of non-tomographic phase sensitive x-ray imaging with breast specimens indicated enhanced visualization of edges and microcalcifications, and correlation with histopathology (Pisano et al., 2000, Stambanoni et al., 2011, Olivo et al., 2013). In vivo phase-contrast mammography using propagation-based technique with synchrotron source have shown improved image quality for evaluating abnormalities and the ability to distinguish benign and malignant abnormalities (Castelli et al., 2011, Longo et al., 2014). In vivo propagation-based phase contrast mammography using a conventional x-ray tube has also been reported (Morita et al., 2008). Topical reviews of various approaches for phase contrast imaging of the breast have been provided (Keyrilainen et al., 2010, Auweter et al., 2014). Task-specific theoretical analysis has shown that the benefit is more likely for diagnostic imaging tasks (Fredenberg et al., 2012). Hence, we focused on a $9.6 \text{ cm} \times 5 \text{ cm}$ field-of-view (FOV) that is approximately similar to small-field mammography used for spot-compression imaging and stereotactic localization (Vedantham et al., 2000b).

II. A. Imaging geometry

The imaging geometry (Fig. 1) was designed to conform to a π -phase grating, 1st order Talbot interferometer operating at design energy of 25 keV in spherical wave (cone-beam) geometry. Emphasis was placed on a compact imaging setup with a source-to-detector distance (SDD=70 cm) similar to digital mammography, while satisfying the requirements for Talbot interferometry. The x-ray incident surfaces of the G0 and G1 gratings were fixed and located at 45 and 650 mm from the x-ray focal spot, resulting in $l = 605 \text{ mm}$. The x-ray focal spot to G0 distance was selected to approximate a mammography x-ray tube (M113 insert with B110 housing, Varian Medical Systems, Salt Lake City, UT). A relatively large value for l was chosen so that the distance d between the G1 and G2 gratings is small and hence, the entire detector and grating assembly comprising the breast support, the G1 and G2 gratings, and the detector, is as thin as possible. This is based on practical considerations for patient positioning with mammography. For parallel-beam geometry with a G1 grating of pitch $p_1 = 4 \text{ }\mu\text{m}$, the first fractional Talbot distance occurs at 40 mm and the pitch of the G2 grating should be $p_2 = 2 \text{ }\mu\text{m}$. For the practical cone-beam geometry (spherical wave), the spacing d between G1 and G2 gratings was corrected (Weitkamp et al., 2005), resulting in $d = 42.8 \text{ mm}$. For the cone-beam geometric parameters ($l = 605 \text{ mm}$, $d = 42.8 \text{ mm}$, and $p_1 = 4 \text{ }\mu\text{m}$) used in the study, the pitch of the analyzer grating (G2), $p_2 = 2.14 \text{ }\mu\text{m}$, and the pitch of the source grating (G0), $p_0 = 30.25 \text{ }\mu\text{m}$. Specifications of the gratings for the cone beam geometry are provided in Table 1. A point x-ray focal spot was used in Monte Carlo simulations. Although the use of a point source negates the need for the G0 grating, the G0 grating was included in these simulations so that x-ray scatter originating from this grating will also be accounted. The breast was modeled as a half-cylinder of 10.5 cm radius (Boone et al., 2000) with elemental composition for adipose or fibroglandular tissue (Hammerstein et al., 1979) and its thickness was varied from 2 to 8 cm. For all breast thicknesses, the exit surface of the breast was located at 645 mm from the x-ray focal spot. The detector was modeled as an ideal detector with pixel pitch of 0.1 mm (Vedantham et al., 2000a, Suryanarayanan et al., 2004) and field of view of 9.6 cm in the direction parallel to the chest-wall (x -direction) and 5 cm in the direction from chest-wall to nipple (y -direction).

The x-ray focal spot, gratings and the detector were centered laterally with respect to the breast and aligned with the chest-wall.

Fabrication of gratings with aspect ratio, defined as the ratio trench-depth: trench-width, as high as 100:1 have been reported (Kenntner et al., 2012, Mohr et al., 2012, Willner et al., 2013). We considered a G2 grating with aspect ratio of 32:1. For the 25 keV design energy, the 32 μm Au-filled trenches of the G2 grating attenuates more than 93% of the x-ray photons. For the imaging geometry, the trench ratio of the G0 grating ($r_0 = 0.75$) defined as the ratio of trench width to pitch (w_0/p_0) was chosen to satisfy the spatial coherence requirement (Pfeiffer et al., 2006). Fabrication of a 9.6 cm $\times$ 5 cm grating would require a 6-inch (150 mm) silicon wafer, which is typically 675 μm thick (Goldstein and Watanabe, 2008). For G0 grating positioned close to the x-ray focal spot, its dimensions can be smaller and can be fabricated from a smaller diameter and a thinner silicon wafer. For simplicity, we assumed all gratings were fabricated from 675 μm thick silicon wafers.

II. B. Monte Carlo simulations

MC simulations of radiation transport were performed by adapting a previous implementation (Vedantham et al., 2012) of a C++ program using the GEANT4 toolkit (version 9.3) (Agostinelli et al., 2003). The Livermore low energy physics model was used. Physical processes for x-ray photon interactions included Compton scattering, Rayleigh scattering and photoelectric effect. All simulations were performed using 25-keV mono-energy x-ray photons, corresponding to the design energy. The simulations were validated by comparing MC estimated transmission efficiencies of the gratings in parallel-beam geometry ($l = 605$ mm, $d = 40$ mm, $p_1 = 4\mu\text{m}$, $p_2 = 2\mu\text{m}$) to match the theoretical computations that were for parallel-beam geometry. For validation, the trench ratio of G0 grating r_0 was varied from 0.1 to 0.9, with $r_1 = r_2 = 0.5$. Upon validation, MC simulations in cone-beam geometry were conducted to determine the transmission efficiencies of the gratings.

II. C. Detector location-independent analysis

For x-ray scatter estimation, each x-ray photon reaching the detector was classified based on interaction along the x-ray beam path as incoherent scatter (S_{incoh}) if it underwent at least one Compton interaction, as coherent scatter (S_{coh}) if it underwent Rayleigh interaction alone, and as primary photons (P) if it satisfied the following 3 criteria: (1) it did not undergo Compton or Rayleigh interaction; (2) its energy matched the incident x-ray photon energy; and, (3) there is no directional change as measured by momentum vectors. Thus, any characteristic x-rays generated due to photoelectric interaction and reaching the detector were ignored in this analysis, unless they underwent at least one scattering interaction. The total x-ray scatter was determined as: $S = S_{incoh} + S_{coh}$. Scatter-to-primary ratio was computed without consideration for detector pixel location (x, y) and is denoted by: $\Sigma S/\Sigma P$, where $\sum S = \sum_x \sum_y S(x, y)$ and $\sum P = \sum_x \sum_y P(x, y)$, respectively. Similar methodology was used to calculate $\Sigma S_{coh}/\Sigma P$ and $\Sigma S_{incoh}/\Sigma P$. Thus, the method treats the detector as a single pixel with dimensions of 9.6 cm $\times$ 5 cm. The number of x-ray photons emitted by the source was varied depending on breast thickness and composition, while targeting $\sim 10 \times 10^5$

photons incident on the detector. This choice provided a coefficient of variation (COV) of 0.75% or less for the primary and scatter fractions determined from three MC runs with an 8 cm thick fibroglandular breast in the x-ray beam.

II. D. Detector location-dependent analysis

For location-dependent analysis of x-ray scatter, an 8 cm thick fibroglandular breast in the x-ray beam was considered as it was likely to provide the highest scatter-to-primary ratio for the range of breast thicknesses studied. MC simulation with three runs of 1.7×10^9 x-ray photons each was used to obtain detector (100 μm pixels) location-dependent estimates of scatter and primary x-ray photons. To improve the statistics in the analysis, the primary and scattered photons were summed along the length of the grating trenches (y - direction in Fig. 1) to provide $P(x)$, $S_{coh}(x)$, and $S_{incoh}(x)$.

II. E. Angular deviation of scattered photons

For each scattered x-ray photon incident on the detector, the momentum vector upon emission from the focal spot, represented by $\vec{F}_e$ and the momentum vector when incident on the detector $\vec{S}_d$ were recorded. The scattering angle θ_s , of each scattered photon was

determined as: $\cos\theta_s = \frac{\vec{F}_e \cdot \vec{S}_d}{\|\vec{F}_e\| \|\vec{S}_d\|}$. The recorded scattering angles were further classified into incoherent scatter and coherent scatter based on the interaction along the x-ray beam path and their distributions were analyzed.

II. F. Poly-energetic x-ray spectrum

The advantage of Talbot-Lau interferometry is the ability to use conventional x-ray tubes that provide poly-energetic spectra. The selection of appropriate kV and filtration, including the use of K-edge filters, can provide for x-ray spectrum matched to the design energy. As an example, we considered a 35 kV W-anode x-ray spectrum filtered by 0.5 mm of Be and 1.75 mm of Al that provided a mean energy matched to the design energy of 25 keV (Poludniowski et al., 2009) and was used in a MC simulation with 8 cm thick fibroglandular breast in the x-ray beam. The results from the simulation were analyzed in a manner similar to that described in Section II. C.

III. Results

Fig. 2 shows the validation of the MC simulation setup under parallel-beam geometry and without the breast in the x-ray beam path. In Fig. 2A, the theoretically computed transmission efficiency for each grating (dashed line) is compared with the MC estimates from 3 runs (symbols), wherein the trench ratio of the G0 grating is varied, while that of the G1 and G2 gratings are constant. Fig. 2B shows the plot of theoretically estimated transmission efficiency vs. MC estimations (symbols). The linear regression fit to the data points and the 95% confidence interval (solid lines) are shown. The dashed line represents identity. The near-unity slope and the near-zero intercept indicate good agreement between theory and MC simulations. Also, the 95% confidence intervals of the slope and intercept span 1 and 0, respectively. Transmission efficiencies of the gratings determined from 4 MC

runs using the cone-beam geometry are summarized in Table 2. Compared to parallel-beam geometry, cone-beam geometry reduced the transmission efficiencies of Au-filled G0 and G2 gratings due to oblique x-ray incidence, while that of G1 is relatively unchanged.

Fig. 3 shows the $\Sigma S_{incoh}/\Sigma P$ (top row: A and B), $\Sigma S_{coh}/\Sigma P$ (middle row: C and D), and $\Sigma S/\Sigma P$ (bottom row: E and F) for adipose (left column: A, C and E) and fibroglandular (right column: B, D and F) breasts. Although G2 grating is essential to facilitate imaging with current detectors, results from simulations performed without the G2 grating (circles in all panels) are provided so that the effect of G2 grating alone can be determined. Results from simulations performed without the breast in the x-ray beam are shown as that for a 0 cm thick breast. Thus, the combined contribution from the source and phase gratings to the ratios can be inferred from the circle at 0 cm in each panel (marked by dashed line labeled ϵ), and that from all three gratings from the square at 0 cm (marked by dashed line labeled τ) in each panel. In panels E and F, the breast thickness above which the G2 grating reduces $\Sigma S/\Sigma P$ is marked by arrow labeled “1”.

Fig. 4A shows the coherent-to-incoherent scatter ratio computed as $\Sigma S_{coh}/\Sigma S_{incoh}$ for 2 to 8 cm thick breasts. For adipose and fibroglandular breasts this ratio ranged from 2.2 to 3.2 and 2.6 to 3.5, respectively, indicating that coherent scatter was the dominant contributor to the total scatter. Fig. 4B shows the percent reduction in incoherent scatter due to the G2 grating

that was computed as
$$\frac{(\sum S_{incoh}/\sum P)^{-G2} - (\sum S_{incoh}/\sum P)^{+G2}}{(\sum S_{incoh}/\sum P)^{-G2}} \times 100$$
, where the superscripts $-G2$ and $+G2$ indicate the estimates obtained without and with the G2 grating, respectively. The percent reduction in incoherent scatter increases with increasing compressed breast thickness in a non-linear manner with ~18% to ~35% reduction for 2 to 8 cm thick compressed breasts.

The primary and scatter distributions along the x -direction after integrating along the y -direction and obtained with 8 cm thick fibroglandular breast are shown in Fig. 5. A periodic pattern is observed in (A), but is not readily apparent in (B). Fourier transform of the data did not show any peaks for scattered photons, but showed peaks at 2.13 and 4.25 cycles/mm for primary photons that were due to object magnification (M) of the source grating G0, as

the spatial frequencies correspond to integral multiples of $\frac{1}{p_0 M}$, where $p_0 = 30.25 \mu\text{m}$ and $M = \frac{SDD}{d_{f-G0}} = 15.56$. This was verified with MC simulations with only the G0 grating in the beam path.

Fig. 6 shows scattering-angle distribution for (A) incoherent and coherent scatter, and (B) coherent-to-incoherent scatter ratio obtained with an 8 cm thick fibroglandular breast. In Fig 6(A), the maxima for incoherent and coherent scattered photons occurred at scattering angles of 0.3125 radians (17.9°) and 0.1125 radians (6.5°), respectively. In Fig. 6(B), the horizontal line at $y = 1$, indicates equal contribution from coherent and incoherent scatter. For scattering angle less than 0.525 radians (30.1°), coherent scatter is the dominant contributor to the total scatter. Over all breast thickness and composition, the percentage of

x-ray photons scattered by an angle $\theta_s > \tan^{-1} \left(\frac{p_2}{d} \right)$ was greater than 99.99% for coherent scatter and greater than 99.96% for incoherent scatter indicating that the reported values are for large-angle x-ray scatter.

Results from simulations using a 35 kV x-ray spectrum with mean energy of 25 keV and with an 8 cm thick fibroglandular breast in the x-ray beam were compared with the corresponding results from simulations using mono-energy x-ray photons of 25 keV. While the $\Sigma S/\Sigma P$ for simulations with mono-energy photons (0.279) and with the poly-energetic x-ray spectrum (0.284) were similar, the $\Sigma S_{incoh}/\Sigma P$ was higher with the poly-energetic x-ray spectrum (0.102) compared to mono-energy x-ray photos (0.07), and the $\Sigma S_{coh}/\Sigma P$ was lower with the poly-energetic x-ray spectrum (0.182) compared to mono-energy x-ray photos (0.204). This suggests that optimizing the x-ray spectrum may be beneficial in reducing the contribution from incoherent scattered photons in acquired images.

IV. Discussion

For the practical cone-beam geometry, data summarized in Table 2 show the need for improving the transmission efficiency of the gratings, in particular G1 and G2 as they are located downstream of the breast. The product of the transmission efficiencies of G1 and G2 is ~17% and is lower than that of anti-scatter grids used in conventional mammography (Chan et al., 1985, Fahrig et al., 1994, Rezentes et al., 1999, Boone et al., 2002). However, this can be substantially improved. For 25-keV x-ray photons, theory indicates ~30% attenuation by the 675 μm thick silicon wafer of each grating, contributing to the reduction in its transmission efficiency. The use of thinner wafers during fabrication of gratings would be beneficial in improving the transmission efficiency. Back-thinning of gratings made from 10 cm diameter wafers down to approximately 220 μm have already been reported (Rutishauser et al., 2013). Spherical or bent gratings would be beneficial in improving the efficiency in cone-beam geometry (Yashiro et al., 2008, Thuring et al., 2013). Improvement in fringe visibility and decrease in uncertainties in the measurement for x-rays incident at angles away from the optical axis with bent gratings have been reported (Revol et al., 2011). The use of spherical gratings matched to radii of curvature in cone-beam geometry is likely to result in transmission efficiencies approaching that of planar gratings in parallel-beam geometry. Back thinning of G1 and G2 spherical gratings to 220 μm could provide for combined transmission efficiency (product of G1 and G2) of ~42%. In screening mammography, the mean glandular dose (MGD) from a single craniocaudal view of an “average” breast or breast-equivalent phantom is limited to 3 mGy. For diagnostic evaluation, the average MGD is approximately 10.2 to 12.4 mGy and the average number of views ranges from 3.65 to 4.53 (Law and Faulkner, 2006, Vedantham et al., 2013). This study focused on a system with a FOV that is suited for diagnostic evaluation. If x-ray phase contrast imaging can be performed at MGD comparable to diagnostic mammography and improves diagnosis then it would be of benefit. A complete analysis should consider task-specific imaging performance (Fredenberg et al., 2012), which is beyond the scope of this manuscript.

Referring to Figs. 3A and 3B, it is observed that the G2 grating acts in a manner similar to an anti-scatter grid in reducing the incoherent scatter-to-primary ratio, $\Sigma S_{incoh}/\Sigma P$. This reduction is observed for the entire range of breast thicknesses considered (2 to 8 cm), and increases with increasing breast thickness. However, in terms of coherent scatter-to-primary ratio, $\Sigma S_{coh}/\Sigma P$ shown in Figs. 3C and 3D, the gratings contributed to an increase (corresponding to τ), where most of the contribution was from G2 (corresponding to $\tau-\epsilon$). Considering that the G2 grating is located very close to the detector, the likelihood that it would contribute to signal in neighboring pixels is relatively low. In these simulations, the scattering angles for coherent scattered photons arising only from G2 were not tracked and hence signal contribution to neighboring pixels could not be determined. The narrowing of the gap in terms of $\Sigma S_{coh}/\Sigma P$ with increasing breast thickness between simulations performed with and without the G2 grating indicates preferential attenuation of coherent scatter compared to primary photons by the G2 grating. Since $\Sigma S = \Sigma(S_{incoh} + S_{coh})$, Figs. 3E and 3F show that for compressed breast thickness greater than approximately 5 cm and 4 cm for adipose and fibroglandular breasts, respectively, the G2 grating reduces the total scatter.

The study observed that large-angle scatter was present for the range of compressed breast thickness encountered in mammography and for a relatively small field of view. The large-angle scatter will manifest as a spatially random field in the acquired image reducing contrast in the transmission image and could contribute to increased statistical uncertainty or noise (Revol et al., 2010) in transmission, dark-field and differential phase contrast images. It is also possible that the scatter-to-primary ratio changes at different G2 grating positions during the acquisition of the multiple images required by the technique. The study utilized homogenous breasts of 100% adipose and fibroglandular tissue composition. For heterogeneous breast, the distribution of tissue structure (Hasnah et al., 2002, Oltulu et al., 2003, Fernandez et al., 2008) will alter the spatial distribution of scatter, in particular coherent scatter. While small-angle and ultra-small angle x-ray scatter can provide valuable information, this study investigated large-angle x-ray scatter. The MC estimated scatter and the above observations of potential effects need experimental validation.

The Livermore low energy physics model of the GEANT4 toolkit uses the EPDL97 data library. Depending on the physics model and data library used there could be differences in interaction cross-sections. Values for β used in some simulation programs may be based only photoelectric cross-sections (del Rio and Dejus, 2003). Also, depending on the model part or all of coherent scatter may be included in δ (Henke et al., 1993). We are aware of one report (Wang et al., 2009) that implemented Monte Carlo-based x-ray refraction by modeling x-ray photon transport in a manner similar to optical photons, but did not investigate large-angle x-ray scatter.

V. CONCLUSION

This study investigated large-angle x-ray scatter in gratings-based Talbot-Lau interferometry for small field of view differential phase contrast imaging of the breast. Emphasis was placed on a compact imaging geometry. Methods to improve the transmission efficiency of the phase (G1) and the analyzer (G2) gratings such as the use of thin, spherical gratings

would be beneficial in reducing the radiation dose. The analyzer (G2) grating reduced incoherent scatter-to-primary ratio incident on the detector for the entire 2 to 8 cm range of breast thickness for adipose and fibroglandular breasts. Coherent scatter was the dominant contributor to the total scatter with coherent-to-incoherent scatter ratio in the range of 2.2 to 3.4. There is active ongoing research on the potential of x-ray phase-contrast breast imaging and this study adds to our understanding of large-angle x-ray scatter.

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Fig. 1.

Imaging geometry of a π -phase grating, 1st order Talbot interferometer operating at a design energy of 25 keV used in Monte Carlo simulations (not drawn to scale). Emphasis was placed on a compact setup, while satisfying the requirements for Talbot interferometry ($l = 605$ mm; $d = 42.8$ mm; $SDD = 700$ mm). The distance between the x-ray focal spot and G0 grating, $d_{f-G0} = 45$ mm. The detector field of view was $9.6 \text{ cm} \times 5 \text{ cm}$ and pixel pitch was 0.1 mm . The breast was modeled as a half cylinder of 10.5 cm radius and its thickness was varied from 2 to 8 cm . For all breast thicknesses, the exit surface of the breast was located at 645 mm from the focal spot.

Fig. 2.

Validation of the Monte Carlo simulations using parallel-beam imaging geometry is shown. The trench ratio of G0 grating was varied, while the trench ratios for G1 and G2 gratings were fixed at 0.5. All gratings were from 675 μm thick silicon wafers. (A) Theoretically computed transmission efficiency for each grating (dashed lines) are compared with results from 3 MC simulation runs (symbols). The symbols overlap between runs as differences were observed only in 3rd decimal place. (B) Linear regression analysis of theory and Monte Carlo simulations. The parameter estimates of slope and intercept are reported as mean $\pm$ standard error, and upper and lower bounds of the 95% confidence interval in square brackets. The 95% confidence interval of the estimates of slope and intercept span 1 and 0, respectively, indicating good agreement between theory and MC simulations.

Fig. 3.

Scatter-to-primary ratios for adipose (left column: A, C and E) and fibroglandular (right column: B, D and F) are shown. The contribution from all three gratings are marked by the horizontal line labeled τ , and that from the source and phase gratings by ϵ . In panels E and F, the breast thickness above which the G2 grating reduces $\Sigma S / \Sigma P$ is marked by arrows labeled "1".

Fig. 4.

The coherent-to-incoherent scatter ratio is shown in (A) and the percent reduction in incoherent scatter due to G2 grating is shown in (B), for 2 to 8 cm thick compressed breasts.

Fig. 5. Spatial distributions of (A) primary photons and (B) scattered photons along the x -direction were obtained by integrating data along the y -direction.

Fig. 6. Angular distributions of (A) incoherent and coherent scattered photons and (B) coherent-to-incoherent scatter ratio with 8 cm thick fibroglandular breast in the x-ray beam path.

Table 1

Specifications of the source (G0), phase or beam-splitter (G1), and analyzer or absorption (G2) gratings. All gratings were considered to be made from 675 μm thick Si wafer substrates. Trench ratio (r) for each grating ($x = 0,1,2$) is defined as the ratio of trench width to pitch.

Description	G0	G1	G2
Pitch, $p_x(\mu\text{m})$	30.25	4	2.14
Trench width, $w_x(\mu\text{m})$	22.69	2	1.07
Trench depth, $d_x(\mu\text{m})$	50	32	32
Trench ratio, $r_x = w_x/p_x$	0.75	0.5	0.5
Trench material	Au	Air	Au

Table 2

Transmission efficiency of the gratings (all 675 μm thick) in cone-beam geometry corresponding to Fig. 1. Transmission efficiency (mean $\pm$ SD) was determined from four Monte Carlo runs without the breast in the x-ray beam.

Grating	Transmission efficiency Mean $\pm$ standard deviation
Source (G0, $r_0 = 0.75$)	$0.158 \pm 2 \times 10^{-4}$
Phase (G1, $r_1 = 0.5$)	$0.695 \pm 6 \times 10^{-4}$
Analyzer (G2, $r_2 = 0.5$)	$0.251 \pm 3 \times 10^{-4}$