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Katie Kern
Lubna Peerzada
Laila Hassan
Carolyn MacDonald

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Katie Kern, Lubna Peerzada, Laila Hassan, and Carolyn MacDonald*

University at Albany, State University of New York, Department of Physics, 1400 Washington Avenue, Albany, New York 12222, United States

Abstract. A system using a wide-slot beam and simple antiscatter grids or slots has been designed to provide a localized map of tissue type that could be overlaid on the simultaneous conventional transmission image to provide an inexpensive, low dose adjunct to conventional screening mammography. Depth information is obtainable from the stereoscopic viewing angles. The system was demonstrated to produce observable contrast between adipose tissue and a phantom chosen to mimic carcinoma at an exposure comparable with screening mammography. Imaging data was collected over a range of system parameters to optimize contrast and to allow verification of simulation modeling. © 2016 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JMI.3.3.030501]

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1 Introduction

Mammography is the gold standard for breast cancer screening, but is known to have both high false-positive and high false-negative rates.^{1–3} Coherent scatter is an alternative method to detect carcinoma within human breast tissue, as cancerous and healthy tissues are known to have different coherent-scatter signatures.^{4–7} For example, the fat scatter peak is due to the average distance between fatty acid molecules of $d = 0.45$ nm, which for the Mo $K\alpha$ characteristic emission at 17.5 keV, corresponds to a peak in the fat scatter at 9 deg. The carcinoma scatter peak is at 13 deg.^{2,5,7–9} The contrast between the tissue types is nearly doubled when using a system which utilizes the characteristic scatter rather than using attenuation imaging.¹⁰ Differences are also seen in collagen alignment, which causes a diminishment of peaks at smaller momentum transfer values.⁵ Coherent-scatter computed tomography (CT) has been used to produce images based on the x-ray scattering properties of the tissue.^{11,12} However, detailed diffraction studies require a small-beam diameter, which would require two-dimensional raster scanning. Similarly, CT imaging is currently incompatible with screening mammography. Utilizing information in the coherent scatter in a system compatible with

screening, i.e., simple, cheap, and low dose, could add entirely new information with the potential to improve specificity and sensitivity.

The design described here uses the intensity of scatter at a fixed, characteristic, coherent scatter angle as a measure of the presence of that tissue type. Unlike the systems used for coherent scatter analysis of tissues, high angular resolution is not required. This and the fact that scatter peaks for amorphous materials such as tissues are inherently fairly broad, allows for the use of large-area beams. Angular selectivity is achieved with a simple focused grid or a slot aperture. The coherent scatter image can be taken simultaneously with a conventional high-resolution transmission image, so high spatial resolution is also not required in the scatter image, allowing low-dose acquisition of tissue typing information. The scatter information is to be used only to highlight areas on the conventional mammo-gram. Thus, there is no tradeoff between conventional and coherent-scatter imaging with this system, as they are achieved simultaneously, with no increase in dose. The goal of this study was to demonstrate that small buried structures could be observed and to provide data to be used to verify a simulation¹³ analysis. The simulation is described in detail elsewhere.^{13–17} Experimental system optimization was performed using signal-to-background ratio (SBR) as the quality parameter. The SBR was defined as

$$\text{SBR} = \frac{I_G - I_F}{I_F}, \quad (1)$$

where I_G is the maximum intensity of the cancer surrogate and I_F is the maximum fat intensity.

2 Experimental Apparatus

2.1 Setup

Coherent scatter is emitted from every volume element of the object at a characteristic angle 2θ . If the beam is apertured to a small area, the cone forms a ring on the detector. An antiscatter grid tilted at that characteristic angle will pass only part of the ring, as shown schematically in Fig. 1.¹⁴ A schematic of the experimental apparatus is shown in Fig. 2.¹³ The shielding creates a slot with a width of 5 mm and height of 60 mm. The slot width is similar to that of clinical slot-scan mammography systems. The low-angle shield was used to block the direct beam and as much of the fat signal as possible. The high-angle shield was used to block the coherent-scatter signal from water in the fat, which occurs at 18 deg. When taking scans of phantoms with a width larger than the slot, the phantom and image plate were moved independently using computer-controlled actuators. In a clinical slot scan system, the source and slot are moved while the patient remains stationary; however, the coherent-scatter collection geometry would be similar. In the setup of Fig. 2, the velocity ratio between the image plate and phantom motion was set equal to the magnification of the system $M = D_{si}/D_{sp}$, where D_{si} is the distance between the source and image plate, and D_{sp} is the distance between the source and phantom. Depth information could be inferred by placing slots and grids on both sides of the direct beam (adding a second setup below the primary beam in Fig. 2). Then the offset due to a change in depth can be distinguished from a

*Address all correspondence to: Carolyn MacDonald, E-mail: c.macdonald@albany.edu

Fig. 1 Coherent scatter with a grid tilted to the characteristic angle. The dashed black arrow is the grid normal. Only the dark part of the ring passes through the grid.

Fig. 2 Schematic of the experimental setup. In the final system, a detector would be placed to capture a conventional mammogram from the primary beam. In this test system, only a single detector was employed. In a clinical slot scan system, the patient (phantom) is fixed and the source and slot are scanned past the patient, but the geometry is otherwise very similar.

lateral offset.^{14,18} The x-ray source used in this experiment was a molybdenum anode fixed tube, Oxford Instruments Apogee, with a maximum voltage of 50 kV, and maximum current of 1 mA. The $K\alpha$ peak for molybdenum is at 17.5 keV, and the $K\beta$ peak at 19.6 keV was suppressed using a zirconium filter. The computed radiography plates used in this experiment were Fuji Imaging Plates Type BAS-III, with dimensions 20 cm $\times$ 25 cm. The image was displayed using ImageGauge software, where red denotes high intensity and blue denotes low intensity; these are converted to black and white for display here. The images were read with a pixel size of 200 μ m.

2.2 Phantoms

The samples chosen for these preliminary measurements were designed to model ductal carcinoma in normal adipose tissue. Frozen bovine fat was used to model normal breast tissue. Freezing did not change the broad diffraction signature from the fat,¹⁹ and the peak position and width from both beef fat and pork fat agreed with a fixed human specimen²⁰ and with literature^{5,12} values.

A phantom was also chosen with a scatter peak close to that of the 13-deg peak of breast carcinoma. In the geometry of this system, the width of the coherent-scatter peak is dominated by the slot size, rather than the inherent material properties, so that a polycrystalline material could be employed as a surrogate for the amorphous tissue. Graphite, with a peak at 12.2 deg and easy machinability, was chosen as the carcinoma phantom. To verify that the graphite material would have similar ring shape and intensity as the amorphous tissues, a measurement of the

Fig. 3 Schematic of four of the phantoms. The fat is shown as beige and the graphite gray. The thickness is out of the plane of the sketch. The buried cube phantom, not shown, is the same as the cube, with two additional 0.75-cm thick fat layers so that the graphite is completely buried in the fat. The rod phantom, not shown, has one piece identical to one of the vertical pieces in the H phantom.

rings in conventional small-beam geometry was performed.¹⁷ The peaks were of similar width, 1 deg for graphite and 1.3 deg for fat. The relative ring width difference will be even less as the aperture is broadened, as it was for the imaging measurements (the slot was 5-mm wide, compared to 3 mm for the ring measurement). The peak height ratio (graphite to fat) was 1.6 and the integrated peak area ratio was 1.3. The peak area ratio is similar to the expectation for carcinoma, as the coherent-scatter cross section for carcinoma is 40% higher than for fat and 20% higher than for glandular tissue.⁷ Additionally, the peak of graphite is closer to the fat peak than is the peak for carcinoma, making the difference in widths of the coherent-scatter peaks less significant as graphite presents a more challenging angle to distinguish. The system is designed to suppress fat and display fibroglandular tissue, with bright areas indicating tumors due to the higher scatter cross section for carcinoma. The inherent scatter contrast is much larger than the attenuation contrast.¹⁰ For the purposes of refining and verifying the system, the goal was to demonstrate the ability to distinguish between fat and the cancer surrogate for small buried samples.

Schematics of the phantoms are shown in Fig. 3. For the checkerboard and cube phantoms, the graphite and fat lie next to each other; there is no overlap. For the “H” and rod phantoms, the graphite sits on top of the fat. The heights vary from 9 to 11 cm, somewhat larger than the slot length of 6 cm. The widths are 4 to 8 cm, again, slightly larger than the maximum scan width of 3.5 cm. Static images of the phantoms show only 5 mm of the width.

2.3 Grids

Two scatter “rejection” grids were used in the experimental system. The first grid had a 10:1 ratio, which results in an acceptance cone angle of 5.7 deg.¹⁸ The dependence of transmission through the grid on the incidence angle is employed to differentiate the coherent-scatter signal. As shown in Fig. 4, if a 10:1 grid is tilted to 14 deg, then graphite is only reduced a little from its peak at 12 deg, but the fat is almost completely blocked. In addition to the grids, lead shields, also shown in Fig. 4, were used to block the low- and high-angle beams. Eventually, it was found that although the 10:1 grid worked as expected to preferentially pass the scatter from the desired material,^{13–15,18} it was very nonuniform in transmission, and so the lead shields were used alone. A second 5:1 grid was more uniform, but had an acceptance angle of 11.3 deg and so was not used to

Fig. 4 Grid transmission of the relevant diffraction peaks for a grid tilted to 14 deg. Gray boxes show the effect of the low- and high-angle shields.

distinguish between coherent-scatter peaks. Trials were made of the effect of placing it perpendicular to the first grid to reduce Compton scatter. It was not found to improve the SBR, and so was removed.

3 Image Formation

The output image from the coherent scatter is a circle, as shown in shown Fig. 5(a) for a small-beam aperture using a solid, 10-cm-thick graphite phantom. If most of the beam is blocked by the low- and high-angle shields, only a partial arc is seen on the image plate. If many small apertures are combined to form a slot, as shown in Fig. 5(b), the partial arcs overlap to form a stripe. The coherent scatter is broader and more intense in the middle, where more arcs from the graphite are added together than near the top and bottom. The coherent-scatter pattern roughly mimics the shape of the target material.

4 Exposure Requirements

Coherent scatter images were obtained from the optimized system with an exposure as low as 100 to 250 mAs with a pixel size of $200\ \mu\text{m}$. The SBR [as defined in Eq. (1)] was nearly independent of exposure time, as shown in Fig. 6, for the checkerboard phantom. This behavior was also true of the other phantom images. Early measurements were taken with longer exposure times because it had been (incorrectly) assumed that longer exposures would increase SBR. In fact, a fast scan with a low exposure time produced the best results, because the image plate tended to self-erase during long exposures. The exposure can be further reduced by employing a detector with larger pixels. For example, if the coherent scatter is used to highlight regions of the original mammogram, it would be sensible to assume that no region less than $1\ \text{mm}^2$ would be useful. Increasing the pixel size in the coherent-scatter image from $200\ \mu\text{m}$ to $1\ \text{mm}$ would result in a reduction in a required dose by a factor of 25. The pixel

Fig. 5 (a) Expanded image of the graphite arc with shields in place, blocking the direct beam and part of the arc. (b) Image of the signal from a large block of graphite.

Fig. 6 Scanned image SBR versus exposure in mAs (times ranging from 2.5 to 12 min, for this low-power source) for the checkerboard phantom with (red squares) and without (blue diamonds) the grid.

size of the final coherent-scatter system would be adjusted so that no additional dose is required beyond that required by the simultaneous conventional mammogram. The plot of Fig. 6 also demonstrates that removal of the grid (to improve image uniformity) slightly decreased the SBR.

5 Effects of Imaging Parameters on Signal-to-Background of Stationary Images

A series of static images were made with different parameter sets to investigate the impact on the SBR. One side of the checkerboard phantom was illuminated with the narrow slot. It was placed first with the graphite on top, then imaged again with the graphite on the bottom, and the SBR from the two images were averaged.

5.1 Detector Distance

The largest increase in SBR was obtained by increasing the sample-to-detector distance from 17.4 to 49.5 cm, with roughly constant source-to-phantom distance of 40 to 42 cm. This allowed greater precision in aligning and adjusting the low- and high-angle shields to block the fat and scatter background. The SBR, for a 5-min exposure with no grid, increased from 1.08 ± 0.05 to 2.65 ± 0.02 .

5.2 Grid Tilt Angle and Low-Angle Shield Location

When a grid was used, the highest signal-to-background (graphite-to-fat ratio) was obtained at 14 deg. This was as expected from the analysis of Fig. 4. The grid was later replaced with simple lead shields because of the nonuniformity issues. The location of the low-angle shield strongly affected the SBR. The SBR increased dramatically when the low-angle shield moved from 10.4 deg to 11.3 deg, blocking the fat background.^{15,17} A further increase in angle did not significantly change the fat background, but began to block the graphite.

5.3 Tube Voltage and Filter Thickness

A Zr filter is required to avoid creating an overlapping diffraction pattern from the $K\beta$ emission. Using a thicker Zr filter suppresses the $K\beta$ but also reduces $K\alpha$. Filter thicknesses of 25, 50 and, $75\ \mu\text{m}$ were compared. The SBR increased by almost 40% when the filter thickness was increased from 25 to $50\ \mu\text{m}$, but did not change between 50 and $75\ \mu\text{m}$. The presence of the filter changes the spectrum and reduces the average photon energy, which might negatively impact patient dose. This can be mitigated by increasing the tube voltage. Measurements were made with fixed tube current and the $50\text{-}\mu\text{m}$ filter, but varying tube voltage. The SBR increased 20% as the tube voltage was increased from 30 to 48 kVp, consistent with the relative

Fig. 7 Images, from left to right, of the checkerboard, cube, buried cube, rod, and “H” phantoms. In all cases, the graphite is visible. The exposure times did not affect the contrasts appreciably and varied from 8 to 12 min. The SBR for the cube and H phantoms were 0.12, and for the rod and buried cube were about 0.08. It is unclear as to why the bottom two graphite pieces of the H phantom are much fainter than the top two graphite pieces. It is possible that the bottom two pieces shifted during the scan, as they were loosely placed within the fat.

increase in characteristic to bremsstrahlung radiation. Neither voltage nor filter thickness has a dramatic effect on the coherent-scatter SBR, and so can be adjusted as necessary to optimize the conventional mammogram. For the scanned images, the tube voltage was set to the maximum allowed by the source, 47.5 kV, largely because the source has very low power, and that minimized the scan time.

5.4 Final Parameter Set and Scanned Image Modulation Transfer Function

The scanned images were taken with no grid, a 49.5-cm sample-to-detector distance, with the low-angle shield located at 10.7 deg, the high-angle shield located at 13.3 deg, and the image plate rotated to 12 deg. The tube voltage was set at 47.5 kV and the current at 0.95 mA (the tube maximum), with a 50- μ m Zr filter.

The velocities of the sample and detector must be the same ratio as that of the magnification. To verify the speed ratios, images were taken of a resolution phantom. Profiles through the image were used to compute contrasts. The static and scanned MTFs, overlapped within measured error,¹⁴ demonstrated that the scanned system was operating properly.

6 Scanned Image Results and Conclusions

A scanned image of the checkerboard phantom, taken with no grid, is shown at the left of Fig. 7. The system has successfully displayed the graphite (cancer surrogate) in the fat matrix. Additional phantom images are shown in Fig. 7. For the H phantom, the central graphite piece is not detected, which, due to the orientation of the system, is expected. Coherent-scatter signal was clearly visible for samples as thin as 2 mm.

After using the measurements to validate the simulation mentioned in Sec. 1, the simulation¹⁷ was used to model carcinoma buried in a 50:50 mixture of fatty and fibroglandular tissue using known scattering parameters for carcinoma.^{21,22} Tumors as small as 0.5 cm were clearly apparent in the simulated images, with an SBR of about 8%. Measurements of actual tumor specimens will be required to fully validate this approach, but the modeling and preliminary phantom measurements indicate that it may provide a useful adjunct to mammography which can be achieved simultaneously, with no additional dose.

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