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Kimberly Kontson
Robert J. Jennings

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Kimberly Kontson^{a,b} and Robert J. Jennings^{a,b,*}

^aCenter for Devices and Radiological Health, Food and Drug Administration, 10933 New Hampshire Avenue, Silver Spring, Maryland 20993, United States

^bUniversity of Maryland, Department of Bioengineering, College Park, Maryland 20742, United States

Abstract. Scatter contamination of projection images in cone-beam computed tomography (CT) degrades the image quality. The use of bowtie filters in dedicated breast CT can decrease this scatter contribution. Three bowtie filter designs that compensate for one or more aspects of the beam-modifying effects due to differences in path length in a projection were studied. These designs have been investigated in terms of their ability to reduce the scatter contamination in projection images acquired in a dedicated breast CT geometry. The scatter magnitude was measured as the scatter-to-primary ratio (SPR) using experimental and Monte Carlo techniques for various breast phantom diameters and tube voltages. The results show that a 55% reduction in the center SPR value could be obtained with the bowtie filter designs. On average, the bowtie filters reduced the center SPR by approximately 18% over all breast diameters. The distribution of the scatter was calculated at a range of different locations to produce scatter distribution maps for all three bowtie filter designs. With the inclusion of the bowtie filters, the scatter distribution was more uniform for all breast diameters. The results of this study will be useful in designing scatter correction methods and understanding the benefits of bowtie filters in dedicated breast CT. © 2014 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.1.3.033505](https://doi.org/10.1117/1.JMI.1.3.033505)]

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1 Introduction

Over the last several years, researchers have been investigating the use of a new imaging modality called dedicated breast computed tomography (CT) as a means of alleviating the problem of tissue superposition that comes from acquiring a two-dimensional image of a three-dimensional object in conventional mammography.^{1,2} Most implementations use dose levels similar to doses in conventional mammography.

The majority of the dedicated breast CT systems undergoing clinical evaluation employ a cone-beam configuration. Several investigators have characterized scatter in clinical cone-beam CT imaging and have concluded that the scatter contributes to image artifacts (such as cupping), increases noise, and decreases the contrast-to-noise ratio (CNR).^{3,4} Given the unique features of dedicated breast CT, the conclusions from scatter studies on general cone-beam CT imaging may not be applicable, so further investigation is needed. A few groups have done such an investigation by looking at the scatter-to-primary ratio (SPR) and scatter profile for different breast compositions, breast diameters, air gap distances, and tube voltages in dedicated breast CT.^{5–7} These studies found that the tube voltage and breast composition had little impact on SPR values. Breast diameter and air gap were the most significant parameters impacting SPR. It was also concluded that the best way to reduce the SPR in the center of the phantom is through the use of a compensator filter, or bowtie filter.⁵

Bowtie filters are designed to deliver a more uniform fluence at the detector in order to reduce the dynamic range requirement on the detector, preferentially harden the x-ray beam, and reduce the dose to the patient.^{8–10} These types of filters are already used in conventional CT scanners. In general, bowtie filters achieve the goal of reducing the dynamic range in the radiation intensity reaching the detector by acting as a complement to or inverse of the patient, i.e., where the ray path through the patient is highly attenuating, the bowtie filter has low attenuation, and vice versa. Due to the lack of symmetry of the human body and to the large differences in the attenuation properties of bone and soft tissue, it is not feasible to design bowtie filters that compensate for the patient in a detailed fashion. In dedicated breast CT systems, however, only the breast is irradiated. Given the homogeneity of the tissue in the breast and its symmetry, a bowtie filter that more precisely compensates for the x-ray characteristics of the breast can be designed.

Our group has designed three different types of bowtie filters that can match certain x-ray characteristics of breast tissue or produce a constant effective attenuation coefficient at the detector.¹¹ Bowtie designs that we have previously referred to as #1 and #2 use the ideas of spectral matching and basis material decomposition either to match the transmitted spectral shape of breast tissue using a single material (bowtie design #1) or to match both the transmitted spectral shape and intensity of transmitted radiation through the breast using two materials (bowtie design #2). For this study, bowtie design #1 was made

*Address all correspondence to: Robert J. Jennings, E-mail: Robert.J.Jennings@fda.hhs.gov

of aluminum and bowtie design #2 was made of beryllium oxide (BeO) and boron carbide (B₄C). Bowtie design #3, which produces the same effective attenuation coefficient at the detector, was made of polymethylmethacrylate (PMMA). Experiments and simulations to evaluate and validate each of the bowtie filter designs are ongoing. Initial experimental and computational results indicate that the bowtie filters accomplish the desired result with regard to the primary (unscattered) beam.

Given the degrading effects of scatter in cone-beam CT images, it is important to understand the consequences and advantages of integrating these bowtie filters into the dedicated breast CT imaging scheme. Several studies have examined the effect of bowtie filters on the distribution and magnitude of x-ray scatter in general cone-beam CT.^{12–14} The results from one study in cone-beam CT using a copper bowtie filter show that the scatter signal is significantly diminished and the horizontal scatter profile is flattened out.¹³ The decrease in the structure of the spatial distribution of the scatter caused by the bowtie filter may be beneficial in certain scatter techniques, where the scatter is assumed to be contained primarily in the low-frequency portion of the projection images.¹³ This report describes our evaluation of the scatter effects of the particular bowtie designs we have developed.

2 Methods and Materials

2.1 Phantoms

2.1.1 Experiment

Hollow cylindrical breast phantoms constructed of PMMA were filled with a liquid simulating the x-ray characteristics of 40% glandular/60% adipose composition breast tissue.¹⁵ Research has shown the average fibroglandular percentage of the female breast to be as low as 20%,¹⁶ while other studies have cited 50% as the average.^{2,17,18} The breast composition of 40% fibroglandular/60% adipose that was used to design the bowtie filters described in this paper falls in between these two extremes. The clinical work by Boone et al.¹⁹ showed that 95% of the uncompressed breast diameters from a cohort of 200 women were between approximately 10 and 18 cm, with the average breast diameter being 14 cm. For this reason, the bowtie filters were designed for a 14-cm diameter cylindrical phantom, and three

phantoms were used with outer diameters of 10, 14, and 18 cm and a PMMA wall thickness of 0.3 cm.

2.1.2 Simulation

Since the liquid mixture used in the physical phantom matched the x-ray characteristics of a breast with the composition based on the Hammerstein data to be within 0.5% in the energy range of interest,^{15,20,21} the phantoms in the simulation were modeled in PENELOPE using the breast composition data from Hammerstein et al.²² The 0.3-cm PMMA shell was also modeled in the simulation geometry. The x-ray source, phantom, and detector were positioned such that the projection of the focal spot through the center of the phantom coincided with the center of the detector.

2.2 Bowtie Filters

The choice of material for each bowtie filter was based upon the availability and the ease of fabrication of the given material. The beam-hardening properties and densities of the materials affect the size and shape of the resulting filters. In order to design a filter that is not too large to fit into clinical scanners or too small to manufacture, a material composed of relatively low atomic number elements and having a high density is desired. Aluminum and PMMA are two materials that produced reasonably sized filters. These two bowtie designs were implemented in both the experimental and simulation analyses. Figure 1 shows the images of bowtie design #1 (Al) and bowtie design #3 (PMMA) that were studied experimentally.

Bowtie design #2 was computationally modeled using BeO and B₄C. These materials are commercially available but are quite expensive and also require specialized fabrication techniques, so an experimental version was not made. These materials were considered because their low atomic number compositions and higher densities resulted in reasonably sized filters that could be experimentally implemented in the future. For this study, the two-material bowtie design #2 filter made of BeO/B₄C was only implemented in simulations. Figure 2(b) shows the cross-sectional profile of this bowtie filter. The cross section for bowtie designs #1 and #3 is shown in Figs. 2(a) and 2(c), respectively.

Fig. 1 Images of the two bowtie filters used for experimental analysis of scatter in dedicated breast computed tomography (CT). (a) Bowtie design #1 made of aluminum that transmits the same quality of radiation as breast tissue. (b) Bowtie design #3 made of polymethylmethacrylate (PMMA) that yields the same effective attenuation coefficient at the detector.

Fig. 2 Cross-sectional views of the bowtie filter designs using (a) aluminum for bowtie design #1, (b) BeO/B₄C for bowtie design #2, and (c) PMMA for bowtie design #3.

2.3 Dedicated Breast Computed Tomography System

2.3.1 Experiment

During a dedicated breast CT scan, the patient lies in the prone position with their breast hanging through a hole in the patient support couch as the x-ray tube and detector assembly rotate around the pendant breast. Our laboratory simulator of this configuration uses a Varian G1582-BI rotating tungsten anode x-ray tube and Varian PaxScan 4030CB detector with a 0.194-mm pixel pitch. Cylindrical phantoms are placed on a rotation stage in the x-ray field. The source-to-isocenter distance and source-to-detector distance were held constant for all experiments at 73.5 and 98.5 cm, respectively. A schematic of the geometry in the xy -plane is shown in Fig. 3.

The bowtie filter designs of all three types were based on the 14-cm phantom. This was based on the assumption that only a single bowtie would be employed in a clinical system. For the 10- and 18-cm phantoms, the source-to-filter distance (SFD) was changed so that the projection of the edges of the bowtie filter matched up with the projection of the periphery of the breast phantom. Previous studies investigating scatter with bowtie filters have also used the technique of varying the SFD to account for various diameter phantoms,^{5,23} and we believe

Fig. 3 Schematic of system geometry for simulation and experimentation. The source is at position (0,0) with the isocenter of the phantom and detector located at 73.5 and 98.5 cm from the source, respectively.

this would be a feasible method to implement in clinical systems. The SFD for each diameter phantom was analytically determined using an equation to describe an x-ray path tangent to the given breast phantom [Eq. (1)]. In this equation, x is the horizontal distance from the source, y is the distance from the source along the source–detector axis, and θ_{tangent} is the angle tangent to the breast phantom (Fig. 3):

$$y = \frac{x}{\tan \theta_{\text{tangent}}} \quad (1)$$

Since the x -coordinate of the peripheral point of the bowtie filter design was known from its nominal position, this value could simply be plugged into Eq. (1) to determine the required translation along the source–detector axis necessary to align the edges of the filter with the periphery of the breast phantom. Table 1 shows the SFDs for three breast diameters.

2.3.2 Simulation

The simulation geometry was meant to match the experimental conditions, so that a direct comparison between the SPR measurements could be made. The half cone-angle and fan-angle of the x-ray beam used in the simulation were 9.29 deg and 8.29 deg, respectively, to irradiate the entire breast phantom for all diameters. Just as in the experimental conditions, the bowtie SFD was adjusted depending on the diameter of the breast phantom. The Monte Carlo simulations used the PENELOPE code²⁴ with the penEasy_Imaging main program.²⁵ This program uses quadrics to represent the geometry in the simulation. A single parabola could not capture the unique curvature of the bowtie filters, so multiple parabolas were fit to multiple segments of the bowtie filter. For each projection image, 10¹⁰ photon histories were simulated using a message

Table 1 The source-to-filter distance (SFD) for three different breast diameters.

Breast diameter	Source-to-filter distance
10 cm	28.4 cm
14 cm	16.25 cm
18 cm	14.9 cm

passing interface code implemented on a 207 node cluster. Each simulated projection took approximately 1350 s. Since the 30 kVp spectrum was so heavily attenuated with bowtie design #1 (A1), 10^{11} photon histories were used to increase the number of photons reaching the detector.

2.4 X-Ray Source and Spectrum

Our computational analysis of the optimal tube voltage for dedicated breast CT indicates that the lower tube voltages will provide the best CNR while also maintaining appropriate dose levels.²⁶ The Varian G1582BI x-ray tube, in combination with our CPI Indico x-ray generator, has a lower kVp limit of 40; therefore, tube voltages of 40, 50, and 60 kVp were investigated experimentally. In the simulations, a tube voltage of 30 kVp was also investigated since tube output is not an issue in the simulation environment. Starting spectra were taken from IPEM Report 78 assuming a 10-deg tungsten anode for each tube voltage (Fig. 4). Each simulated spectrum was filtered with 0.8 mm of beryllium and 1 mm of aluminum to match the inherent filtration in the x-ray tube. A 0.1-mm copper filter was also placed in the x-ray beam.

2.5 Scatter-to-Primary Ratio and Scatter Distribution

The SPR was experimentally measured in the x -direction using the lead strip beam block method.^{5,27,28} Since the initial bowtie filters are not designed to account for variations in the z -direction, only the SPR in the x -direction was measured. A rectangular lead strip with dimensions 2 mm $\times$ 250 mm $\times$ 3 mm thick was sandwiched between two thin sheets of PMMA and placed in the x-ray beam between the bowtie filter and phantom (Fig. 5).

Two sets of images were acquired: one with the lead strip and one without the lead strip. These projection images were dark-field corrected, and regions of interest (ROIs) in the projection of the lead beam block in both images were averaged and analyzed to determine the SPR as a function of distance across the detector according to the following equation:

Fig. 4 Number of simulated photons per energy bin for each simulated tube voltage.

**Projection image with lead strip beam block
(x -direction)**

Fig. 5 Experimental projection image using the lead strip beam block method.

$$\text{SPR}(x) = \frac{\xi_S(x)}{\xi_{P+S}(x) - \xi_S(x)}.$$

In this equation, x is the horizontal distance across the detector, ξ_S is the grayscale pixel value of the beam block projection image, and ξ_{P+S} is the grayscale pixel value of the primary + scatter image without the beam block.

The Monte Carlo code PENELOPE is capable of generating images without scatter, with scatter, and also images from individual types of scatter events (i.e., Compton, Rayleigh, and so on). Using these images, the SPR can easily be calculated as a function of distance across the detector. To match the experimental methodologies, however, a lead beam block strip with the same dimensions as the strip used experimentally was included in the geometry of the simulation in the x -direction. The SPR was then calculated according to Eq. (2), using the same size ROIs.

Information on the scatter distribution was obtained using the simulation images. Bootsma et al.¹³ introduced a scatter ratio relationship that helps to highlight subtleties in the scatter signal that might otherwise be overwhelmed by the primary signal in the SPR. The scatter-to-open-field center pixel ratio (SOCCR) looks at the ratio of the scatter only signal (S) and the central pixel value in the flat-field image (I_{open}^0), where no object attenuation occurs [Eq. (3)]. To compute this ratio, the scatter signal image was rebinned from a 1536×2048 image to a 48×64 image such that the scatter signal was averaged over 6.2 mm^2 blocks. In Eq. (3), x' and y' represent the bin number of the rebinned images:

$$\text{SOCCR}(x', y') = \frac{S(x', y')}{I_{\text{open}}^0}. \quad (3)$$

The ratio was computed for all diameter phantoms for all three bowtie filters and no-bowtie filter cases for the 50 kVp simulation. The resulting bowtie/no bowtie images that correspond to

Fig. 6 Experimental scatter-to-primary ratio (SPR) values for various tube voltages using (a) no bowtie filter, (b) bowtie design #1, and (c) bowtie design #3. All results shown here are for the 10-cm diameter phantom.

the projection of the cylindrical phantom were shown on the same scale.

3 Results

3.1 Scatter-to-Primary Ratio Dependency on Tube Voltage

The SPR was calculated for three different tube voltages in the experiments, and for four different tube voltages in the

simulations. Figure 6 shows the experimental SPR values obtained with the 10-cm breast phantom with and without bowtie design #1 and bowtie design #3. The 10-cm phantom results were used because the limited tube output did not allow for accurate SPR measurements for the 14- and 18-cm diameter phantoms at low tube voltages.

Figure 7 shows the SPR calculations from the simulated images for the 10-cm breast phantom as a function of tube voltage.

Fig. 7 SPR calculations from simulated images for various tube voltages using (a) no bowtie filter, (b) bowtie design #1 (Al), (c) bowtie design #2 (BeO/B₄C), and (d) bowtie design #3 (PMMA). All results shown here are for the 10-cm diameter phantom.

Fig. 8 Calculated percent change from center SPR value (at $x = 0$) with no bowtie filter and center SPR value with bowtie filter for various tube voltages for 10-cm diameter phantom.

Both the experimental and simulation results show that for higher tube voltages, the SPR does not drastically change. However, as the tube voltage decreases, the SPR tends to increase. This same trend was seen in the experimental and simulation work characterizing the scatter in dedicated breast CT by Chen et al.⁶ With the use of the bowtie filters, the SPR can be decreased even with the lower tube voltages. Figure 8 shows the percent decrease from the SPR value at the center of the phantom ($x = 0$) with no bowtie filter to the center SPR value (at $x = 0$) with the three different bowtie filters from the simulation data. The greatest reduction in the SPR value compared with the no-bowtie case was obtained at the lower tube voltages. The center SPR value was reduced the most with bowtie design #2 (BeO/B₄C) for all tube voltages, reaching a maximum reduction of 55% with the 30-kVp spectrum.

3.2 Scatter-to-Primary Ratio Dependency on Breast Diameter

The bowtie filters were designed for a specific breast diameter. From initial computational results analyzing the robustness of the bowtie filter designs, breast diameter did not change the desired result greatly. The following plots show how the SPR is affected by varying breast diameters with the use of the three bowtie filters.

Figure 9 shows the SPR calculated through experimental images and simulation images for all bowtie designs and for three different breast diameters at 50 kVp. There is no physical prototype of bowtie design #2 (BeO/B₄C) so only simulation data are available. The experimental and simulation data match for the larger diameter breast phantoms, but deviate by approximately -17% for the 10-cm phantom. For each breast diameter, the use of the bowtie filter decreases the SPR value at the center and also flattens the SPR value in the x -direction.

Both bowtie designs #1 (Al) and #3 (PMMA) exhibit similar behavior for all three breast diameters. According to the simulation data, the overall magnitude of the scatter (maximum in the SPR curve) is decreased for each breast diameter by an average of approximately 18% by these two bowtie filter designs.

The largest reduction in SPR of approximately 35% comes from using bowtie design #2 (BeO/B₄C). For this bowtie design, the SPR as a function of distance across the detector is nearly flat for the 14-cm phantom. With the 10-cm phantom, bowtie design #2 produces cupping in the SPR compared with the 14-cm plot: the lowest value is located in the center while the edges of the phantom exhibit large SPR values. A similar result was found by Kwan et al.⁵ in their characterization of scatter properties in dedicated breast CT. The bowtie filter used in the Kwan study was made of Teflon (C₂F₄) and designed for a 14-cm cylindrical phantom made of 50% fibroglandular/50% adipose composition at 80 kVp. Their results show the SPR at the center of the 10-cm phantom to be approximately 0.23, increasing to a value of approximately 0.6 at the periphery. The results in the present study show an SPR value of 0.28 at the center of the 10-cm phantom and approximately 0.7 at the periphery. With the larger breast diameter, bowtie design #2 causes capping in the SPR compared with the 14-cm plot. Again, this same result is seen in Kwan et al.

Lück et al.²⁹ recently reported a paper on the effects of bowtie filtration on image quality and dose in dedicated breast CT. This simulation study looked briefly at the characterization of scatter with bowtie filters for different breast diameters. Figure 10 shows the center SPR value for each breast diameter and bowtie filter in the present study as well as the values obtained by Lück et al. The same general trend is seen with the central SPR values calculated using the three bowtie filter designs in this study compared with the bowtie design from Lück et al. A direct comparison is difficult to do, however, since the design goals, bowtie filter materials, and system geometries differ between the two studies.

3.3 Scatter Distribution

Using the ratio given by Eq. (2), the effect of the bowtie filter on the distribution of scatter was analyzed. Each bowtie filter contains a constant thickness base of 0.35 cm of the bowtie material (s). This base thickness was included in the no-bowtie configurations, so the no-bowtie filter case for the scatter distribution was different for each bowtie filter. Figures 11–13 show the scatter distribution for 10-, 14-, and 18-cm diameter phantoms at 50 kVp. The left column shows the no-bowtie filter case with the added thickness base of each bowtie filter material. The right column shows the scatter distribution with the bowtie filter. For each phantom diameter, the images are on the same scale so that the difference between the bowtie/no-bowtie distributions can easily be seen, and a comparison between the distributions for each bowtie filter can be made.

The scatter distributions are very similar for the no-bowtie filter cases for a given breast diameter. With these plots, it is easy to see the reduction in the magnitude of the scatter signal for all three bowtie types. The greatest reduction in the scatter magnitude can be seen using bowtie design #2, as was indicated with the SPR measurements. For the 10-, 14-, and 18-cm diameter phantoms, the inclusion of bowtie design #2 reduced the scatter magnitude by 38%, 38%, and 50%, respectively. As the breast phantom diameter is increased, the structure in the scatter distribution plots also increases with the no-bowtie filter case. With the inclusion of the bowtie filters, it can be seen that the distributions of the scatter in the x - and y -directions are more uniform compared with the no-bowtie filter cases. For example, the maximum and minimum values of the SOCR along the line $y' = 25$ for the 18-cm diameter phantom with no bowtie in

Fig. 9 SPR value for various diameter phantoms using (a–c) bowtie design #1 (Al), (d–f) bowtie design #2 ($\text{BeO/B}_4\text{C}$), and (g–i) bowtie design #3 (PMMA). Both experimental and simulation values of SPR are shown at a tube voltage of 50 kVp. The legend in (b) is valid for all plots.

Fig. 10 Comparison of central SPR values obtained in the present study and those values obtained from Lück et al.²⁹ The central SPR value for the simulation data was plotted for each bowtie filter type and breast diameter.

Fig. 13(c) are 0.0176 and 0.0130 (26% difference). With bowtie design #2, the maximum and minimum values along the same line in Fig. 13(d) are 0.00885 and 0.00821, respectively (7% difference). While this result can be readily seen in the SPR measurement for bowtie design #2 ($\text{BeO/B}_4\text{C}$), the increased uniformity in the scatter distribution with bowtie designs #1 (Al) and #3 (PMMA) is more difficult to see using the SPR metric. With the SOCR metric, the decrease in the structure of the scatter signal is easier to appreciate.

4 Discussion

Through experimental and computational methods, the scatter in dedicated breast CT with the use of three different bowtie filters has been characterized in terms of magnitude and distribution. These filter designs were evaluated to see how they affect the scatter with varying breast diameters and tube voltages. The SPR values were obtained in the x -direction. Since the photons travel through the same filter thickness at the center in the bowtie and no-bowtie filter cases, the SPR results obtained in the y -direction would be similar between the no-bowtie and bowtie filter cases. Bowtie design #2 ($\text{BeO/B}_4\text{C}$) produced the greatest reduction in scatter magnitude and provided a nearly uniform scatter distribution. The more uniform dose distribution was to be expected based on the design: by using two materials,

Fig. 11 Scatter distribution using the scatter-to-open-field center pixel ratio (SOCR) metric for the three bowtie filter types using a 10-cm diameter breast phantom and a tube voltage of 50 kVp. (a), (c), and (e) show the scatter distributions with no bowtie filter, but include the constant thickness base from the bowtie filter so that a valid comparison between the two cases can be made. The variables x' and y' represent the bin number of the rebinned scatter image used to compute the SOCR. (a) No bowtie + Al. (b) Bowtie design #1 (Al). (c) No bowtie + BeO/B₄C. (d) Bowtie design #2 (BeO/B₄C). (e) No bowtie + PMMA. (f) Bowtie design #3 (PMMA).

Fig. 12 Scatter distribution using the SOCR metric for the three bowtie filter types using a 14-cm diameter breast phantom and a tube voltage of 50 kVp. (a), (c), and (e) show the scatter distributions with no bowtie filter, but include the constant thickness base from the bowtie filter so that a valid comparison between the two cases can be made. The variables x' and y' represent the bin number of the rebinned scatter image used to compute the SOCR. (a) No bowtie + Al. (b) Bowtie design #1 (Al). (c) No bowtie + BeO/B₄C. (d) Bowtie design #2 (BeO/B₄C). (e) No bowtie + PMMA. (f) Bowtie design #3 (PMMA).

Fig. 13 Scatter distribution using the SOCR metric for the three bowtie filter types using a 18-cm diameter breast phantom and a tube voltage of 50 kVp. (a), (c), and (e) show the scatter distributions with no bowtie filter, but include the constant thickness base from the bowtie filter so that a valid comparison between the two cases can be made. The variables x' and y' represent the bin number of the rebinned scatter image used to compute the SOCR. (a) No bowtie + Al. (b) Bowtie design #1 (Al). (c) No bowtie + BeO/B₄C. (d) Bowtie design #2 (BeO/B₄C). (e) No bowtie + PMMA. (f) Bowtie design #3 (PMMA).

this design transmits the same intensity and quantity of radiation as breast tissue, making it seem as if the object being imaged is a uniform block.

For different tube voltages, both the experimental and simulation data showed an increase in the SPR value with decreasing tube voltage. A similar trend was reported in previous studies.^{5,6} A wide range of optimal tube voltages has been proposed for dedicated breast CT ranging from low values near 30 kVp to higher values around 80 kVp.^{2,18,26,30,31} It would be advantageous to use a bowtie filter that reduces the scatter independent of the tube voltage.

Given this range. With the inclusion of a bowtie filter, the SPR was decreased for all tube voltages, but this decrease was not completely independent of tube voltage. There was also some variability in the amount of scatter reduction obtained using each bowtie filter at various tube voltages.

The difference seen between bowtie filters in the scatter reduction at different tube voltages (Fig. 8) may be partly due to the amount of scatter generated by the bowtie filter itself. The ratio between the incoherent scatter cross section and the total linear attenuation coefficient for the four bowtie filter materials increases rapidly with increasing energy between 10 and 50 keV.³² Since incoherent scatter contributes the most to the scatter contamination signal reaching the detector, this ratio would help to explain the decrease in scatter reduction seen with higher tube voltages for each bowtie filter design. Future studies could include the analysis of the scatter contribution from each individual component in the CT scanner/simulation environment.^{6,33,34} With these data, more insight into the changes of the SPR at the periphery of the phantoms could be gained. Also, more extensive work could be done to see how the redirected photons affect focal spot size and modulation transfer function.

Figure 6(b) points out a limitation in the current experimental setup. While imaging was used to align the bowtie filter with the periphery of the give breast phantom, this is not the most accurate method of alignment. The bowtie filter during this experiment could have been misaligned along the x -direction (perpendicular to the source–detector axis). This would cause the rays traveling through the peripheral portion of the phantom on one side to be more attenuated since they will all be traveling through the thickest portion of the bowtie filter. This could decrease the amount of primary fluence reaching the detector, causing the increase in the SPR seen in Fig. 6(b).

The experimental and simulation SPR data for the three different bowtie filters with three different breast diameters matched reasonably well. Although there was an approximate –17% difference between the experimental and simulation SPR values for the 10-cm breast phantom, the overall shape of the SPR curves matched very well. This difference in the SPR values at the smaller breast diameters could be due to the larger collimation used in the experimental setup. Future studies will use variable collimation depending on the breast phantom diameter used, since this will provide more consistent results and also reduce the amount of scatter that reaches the detector.

Bowtie filter design #2 (BeO/B₄C) showed the greatest reduction in the SPR for all breast diameters. The SPR as a function of horizontal distance was nearly flat for the 14-cm diameter phantom using this bowtie design. Since this bowtie is designed to attenuate x-rays the same way that 40% glandular/60% adipose breast tissue does, it would seem as if the x-rays all traveled the same path length to get to the detector at all fan angles.

The variation in bowtie filter performance for different breast diameters for bowtie designs #1 (Al) and #3 (PMMA) is very small compared with the variation in bowtie filter performance with bowtie design #2 (BeO/B₄C). The SPR is reduced using bowtie designs #1 and #3 for all breast diameters, and the general shape of the SPR curve remains the same. Although the center SPR value is reduced for all three breast diameters using bowtie design #2 (BeO/B₄C), the shape of the SPR curve changes with the breast diameter. At smaller breast diameters, the SPR curve is higher at the periphery of the phantom and lower at the center. The bowtie filters are designed by taking into account the change in the path length of a photon through a cylindrical phantom as a function of fan-angle. The bowtie filter thickness is minimum at the center, and is the same for the bowtie and no-bowtie filter cases so the effect of the bowtie filter is not applicable. Moving toward the periphery, however, the variation of the bowtie filter is meant to account for the 14-cm phantom, and not the 10-cm phantom. Figure 14 shows the shape of the bowtie filter if it were specifically designed for the 10-cm phantom.

The difference in the thickness of the filter can be easily seen. This means that when imaging a 10-cm phantom with the bowtie filter that was designed for a 14-cm phantom, the photons are being attenuated too much. This results in fewer primary photons hitting the detector at the larger fan-angles, which drives the SPR value higher and higher. The opposite is seen for the 18-cm phantom. The bowtie filter #2 design used in the study is under-compensating for the path length through the 18-cm diameter phantom, which results in more primary photons hitting the detector at larger fan-angles. This explains the decrease in SPR toward the periphery of the phantom with the larger breast diameter.

The last analysis done on the simulation images looked at the spatial distribution of the scatter signal with the three different bowtie filters for three different diameter phantoms and a 50 kVp spectrum. These results reveal more information about the distribution of the scatter signal than the SPR, since

Fig. 14 Comparison of bowtie designs for the 10- and 18-cm diameter phantoms with the design used in the present study (14-cm diameter phantom). The shape of the two-material bowtie filter designed specifically for the 10-cm phantom is shown as the thin solid line, while the specific design for the 18-cm phantom is shown as the thick solid line. The 14-cm phantom design used in the present study is shown as a dotted line.

the SPR values can be overshadowed by the large variation in the primary signal. As indicated by the plots in Figs. 11–13, the bowtie filters reduce the magnitude of the scatter and the spatial variation in the scatter distribution. The benefit of a bowtie filter in reducing the scatter structure is more evident with larger phantoms. Bowtie design #2 (BeO/B₄C) achieves the highest degree of uniformity in the scatter signal distribution. The ability of these bowtie filter designs to produce a more uniform scatter signal distribution could be beneficial when applying scatter correction algorithms that use primary modulation to separate the primary and scatter signals in the frequency domain.^{35,36} This feature could also increase the computational efficiency in Monte Carlo-based scatter corrections that rely on fitting functions to reduce the number of photons required to estimate the scatter.^{37,38}

Dedicated breast CT systems that have been described in the literature generally align the ray that enters the breast closest to the chest wall so that it is parallel with the chest wall, in order to image as much breast tissue as possible. Our setup is similar to that setup, since at the vertical midpoint of the phantom, where our data were taken, the x-ray beam is perpendicular to the axis of the cylinder, and the diameter of the cylinder is the same as the diameter of the breast at the chest wall. In a clinical system, in the part of the breast near the chest wall, there is a scatter contribution from radiation scattered in the direction of the chest wall and then backscattered. In our system, there is the same contribution and in addition a larger contribution from the portion of the phantom above the midpoint since it is also in the x-ray beam. Thus, our data probably approximate the scatter in the clinical geometry near the chest wall but are displaced somewhat from the chest wall distally, i.e., in the posterior-anterior direction. The parameters in this study would be more representative of the clinical situation with an anthropomorphic phantom and a setup consistent with clinical systems. A fully 3-D bowtie filter that accounts for the variation in the breast cross-sectional diameter in the z-direction could also be designed and tested with an anthropomorphic phantom to determine the benefits in scatter reduction. Our current work is focused on the design and evaluation of such a 3-D bowtie filter in both the simulation and experimental environments.

5 Conclusions

Three bowtie filters have been designed to improve the image quality and reduce the dose to the patient in dedicated breast CT. While these bowtie designs have previously been investigated to determine their effect on primary radiation, their effect on scatter has not, until now, been characterized. The present study investigated the scatter magnitude and distribution with and without the three bowtie filters using the SPRs and a newly proposed SOCR, respectively. The results show that the scatter magnitude can be reduced up to 35% with the inclusion of the bowtie filters. Spatial distribution of the scatter is also more uniform with the use of all bowtie filters.

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Kimberly Kontson received her bachelor's degree from the Pennsylvania State University in biomedical engineering. She returned to school after a period of employment with the Maryland Psychiatric Research Center to obtain her PhD degree in bioengineering from the University of Maryland, in 2014. Her dissertation research at the U.S. FDA in the Division of Imaging, Diagnostics, and Software Reliability focused on the development and evaluation of bowtie filters for dedicated breast CT systems.

Robert J. Jennings is a research physicist at the Center for Devices and Radiological Health, FDA. His areas of interest include x-ray physics and x-ray imaging, with current emphasis on new technologies for x-ray imaging of the breast. His duties include computational and experimental research into evaluation methodology for x-ray-based breast imaging devices. His duties also include regulatory review of x-ray imaging devices.