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Observer detection limits for a dedicated SPECT breast imaging system

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Abstract

An observer-based contrast-detail study is performed in an effort to evaluate the limits of object detectability using a dedicated CZT-based breast SPECT imaging system under various imaging conditions. A custom geometric contrast-resolution phantom was developed that can be used for both positive ('hot') and negative contrasts ('cold'). The 3 cm long fillable tubes are arranged in six sectors having equal inner diameters ranging from 1 mm to 6 mm with plastic wall thicknesses of <0.25 mm, on a pitch of twice their inner diameters. Scans of the activity filled tubes using simple circular trajectories are obtained in a 215 mL uniform water filled cylinder, varying the rod:background concentration ratios from 10:1 to 1:10 simulating a large range of biological uptake ratios. The rod phantom is then placed inside a non-uniformly shaped 500 mL breast phantom and scans are again acquired using both simple and complex 3D trajectories for similarly varying contrasts. Summed slice and contiguous multi-slice images are evaluated by five independent readers, identifying the smallest distinguishable rod for each concentration and experimental setup. Linear and quadratic regression is used to compare the resulting contrast-detail curves. Results indicate that in a moderately low-noise 500 mL background, using the SPECT camera having 2.5 mm intrinsic pixels, the mean detectable rod was ~3.4 mm at a 10:1 ratio, degrading to ~5.2 mm with the 2.5:1 concentration ratio. The smallest object detail was observed using a 45° tilted trajectory acquisition. The complex 3D projected sine wave acquisition, however, had the most consistent combined intra- and inter-observer results, making it potentially the best imaging approach for consistent results.

1. Introduction

Nuclear medicine breast imaging is a promising complementary molecular imaging technique to x-ray mammography and computed tomography (CT) because of its ability to provide functional information that can help distinguish malignant from benign tissue. Small concentrations of injected compounds that are ^{99m}Tc radiolabeled are imaged, providing functional information about the metabolic activity of the lesion. To date, there is one FDA-approved compound (^{99m}Tc-sestamibi) for breast cancer imaging. Combining planar x-ray mammography and planar single photon scintimammography (both 2D imaging techniques) has been shown to increase the accuracy of diagnosis of primary breast cancers (Buscombe *et al* 2001). Studies have also indicated that the accuracy of scintimammography alone is

unaffected by breast density and that women with dense breasts may benefit most from this complementary imaging technique (Coover *et al* 2004, Khalkhali *et al* 2002).

Early scintimammography studies using conventional whole-body gamma cameras found a sensitivity and specificity of scintimammography in the detection of malignant breast tumors of approximately 85% (Taillefer 1999, Waxman 1997), but sensitivity dropped significantly for tumors smaller than 1 cm (Tofani *et al* 1999). To overcome this, several groups have investigated higher resolution, small field-of-view cameras for planar scintimammography (Brem *et al* 2005, Hruska *et al* 2008, Spanu *et al* 2007, More *et al* 2006). Given the increased image information content, it is believed that the diagnostic accuracy may further be improved when using 3D tomographic techniques.

Thus, our lab has developed a dedicated single photon emission computed tomography (SPECT) breast imaging system that allows for fully 3D imaging of a hemispherical volume about a pendant breast (Archer *et al* 2003, Tornai *et al* 2003, Brzymialkiewicz *et al* 2005, 2006). The patient lies prone on a customized table, while the system images about her pendant, uncompressed breast (figure 1). The independent SPECT subsystem has recently been coupled with a low dose cone-beam x-ray CT subsystem, also developed in our lab, but CT data were not acquired in this study. The compact SPECT gamma camera and flexible camera gantry allow for innovative 3D imaging trajectories, as opposed to simple circular orbits, using a camera tilt to image a greater volume of the breast and lesions close to the chest wall. Breast imaging with clinical SPECT cameras is limited by the bulkiness of the large whole body cameras, which results in a larger radius of rotation (ROR). Spatial resolution degrades with increasing distance from the collimator in SPECT, and therefore, the larger ROR additionally results in degradation in image quality. Our compact gamma camera and computerized motion system gantry overcome the physical proximity restrictions of clinical gamma cameras, allowing close contouring of the pendant breast while potentially avoiding high-activity background regions outside the breast volume of interest.

With the implementation of this 3D dedicated breast SPECT imaging system, it is necessary to evaluate and characterize system performance to provide tangible motivation for further clinical testing of this paradigm. In previous studies, lesion detectability for varying lesion sizes and contrast ratios has been evaluated using the quantitative signal to noise ratio and lesion to background contrast measurements (Archer *et al* 2003, Brzymialkiewicz *et al* 2005, Tornai *et al* 2005a). However, an observer study is desirable to further characterize the system for an object detection imaging task.

Contrast-detail observer studies are commonly used for comparing varying imaging systems and techniques and have been regularly used for nuclear medicine tomographic and planar imaging systems (Faulkner and Moores 1986, More *et al* 2006). The goal of this initial phantom-based study is to compare multiple 3D SPECT imaging trajectories possible with our system, and evaluate the minimum object size under a variety of 'hot' and 'cold' signal to background contrast ratios since early and later stage, and more and less aggressive cancers take up varying amounts of tracer compounds (Maublant *et al* 1996, Wackers *et al* 1989). Imaging 'hot' objects is appropriate because most cancerous lesions are more metabolically active than the surrounding normal tissue. 'Cold' imaging is also appropriate because, as cancers become more advanced, they may contain necrotic cores that do not uptake or concentrate radioactive tracers. Object size (detail) is also important as early detection of smaller lesions potentially increases survival rates along with patient treatment options.

2. Materials and methods

2.1. Gamma camera

A $16 \times 20 \text{ cm}^2$ area cadmium zinc telluride (CZT)-based compact gamma camera (LumaGEM 3200S, Gamma Medica, Northridge, CA) with $2.3 \times 2.3 \times 5 \text{ mm}$ discrete crystals spaced on a 2.5 mm pitch was used for all imaging studies. The camera has a measured mean energy resolution of 6.7% FWHM at 140 keV and sensitivity of 37.9 cps MBq^{-1} . A low energy collimator with 1.22 mm (flat-to-flat) hexagonal hole openings, 0.2 mm septa and 2.54 cm height was used. The gamma camera is attached to a laboratory jack (model M-EL120, Newport Corp., Irvine, CA) and a goniometer (model BGM200PE, Newport Corp., Irvine, CA), which provide the maneuverability for flexible, 3D trajectories traversing a hemisphere about the pendant breast. The hybrid SPECT-CT system is united on a common gantry (model RV350CCHL, Newport Corp., Irvine, CA), which rotates the system about the breast's central axis (figure 1). The overall performance characteristics and breast imaging capabilities of this specific camera system have been previously described (Brzymialkiewicz *et al* 2005).

2.2. Contrast-detail phantom design

A custom geometric contrast-resolution phantom was developed that can be used for both positive ('hot') and negative contrasts ('cold') (figure 2). The frame was designed using Autodesk Inventor software and then constructed using 3D stereolithography (American Precision Prototyping, Tulsa, OK) using a water-resistant resin (DSM Somos® 11120, density $\sim 1.12 \text{ g cm}^{-3}$). The 3 cm long thin-walled polytetrafluoroethylene (PTFE, density $\sim 2.2 \text{ g cm}^{-3}$) plastic tubes (Small Parts, Inc., Miami Lakes, FL) of varying inner diameters (6.2, 5.0, 4.0, 3.1, 2.3 and 1.2 mm with equal wall thicknesses of $< 0.25 \text{ mm}$), on a pitch of twice their inner diameter were arranged in six sectors. For the three smallest diameter sectors, two additional tubes were added, spaced at three times their inner diameter, in order to evaluate if rods would be visible at all, regardless of satisfying Nyquist's criteria. Tubes were independently filled using equal concentrations of radioactivity and then sealed on their open ends with thin plastic tape.

Though loosely based on a commercially available mini-cold rod phantom we have used in the past (model ECT/DLX-MP, Data Spectrum Corp., Hillsborough, NC), this newly developed phantom can be used for any combination of concentrations, both for hot and cold spot imaging, while also varying the background activity and (fluid density) composition. The thin wall of the tubes is intended to minimize scatter and partial volume sampling effects due to the minimal introduction of a 'cold' region between the tube interstitial space and the remainder of the phantom. The phantom can be used for a variety of other applications, such as for SPECT-CT multi-modality imaging and co-registration, by placing aqueous activity in the tubes and replacing the background liquid with fluids of varying density to modify the CT contrast along with simulated biological uptake of radioactive compounds.

2.3. Contrast-detail observer study

Imaging experiments with geometric and anthropomorphic phantoms were designed. The first set of experiments was designed to optimally image the cylindrical geometric rod phantom using a close ROR of 4.3 cm and low scatter conditions. The phantom was placed in a 7.7 cm inner diameter cylinder, and the background volume was filled with 215 mL of water (figure 2, RIGHT). The tubes were individually filled with aqueous $^{99\text{m}}\text{Tc}$ -pertechnetate radioactivity with an absolute concentration of $10 \mu\text{Ci mL}^{-1}$, approximately five times the lesion uptake found in biodistribution data obtained via pharmaceutical uptake studies (Maublant *et al* 1996, Wackers *et al* 1989). The background concentration was

varied, resulting in tube:background ratios of infinite:1, 10:1, 5:1, 2.5:1, 0.4:1, 0.2:1, to 0.1:1 (with ~5% error due to variations intrinsic to the dose calibrator). The varying contrast ratios in this study are modeled after clinical findings that ^{99m}Tc -sestamibi concentrates in breast tumors with a mean contrast ratio of ~5.6:1 compared to the surrounding normal tissue, varying from ~2.6:1 up to ~8.7:1 (Maublant *et al* 1996).

A vertical axis of rotation (VAOR) 360° circular acquisition orbit was used to acquire a total of 128 projections. Scan times were initially 10 min and subsequently lengthened to account for radioactive decay. The initial count rate was ~300 counts per second (cps) for the 10:1 contrast ratio, thereafter increasing proportionally with the added background activity. Taking advantage of the excellent energy resolution of the semiconductor-based CZT gamma camera, an 8% ($\pm 4\%$) window about the 140 keV photopeak was used for this and all of the following studies.

In the second set of experiments, the contrast-detail phantom was removed from the cylinder and placed inside a non-uniformly shaped breast phantom (Tornai *et al* 2005b), giving more clinically realistic attenuation and scatter characteristics. The dimensions of the phantom were approximately $7 \times 14 \times 17$ cm (nipple-to-chest depth, superior-inferior height, medial-lateral width, respectively). The breast phantom was only partially filled with 500 mL of water in order to avoid paralyzing the detector with the continually added activity to vary concentration ratios, but sufficient to completely submerge the contrast-detail rod phantom. Scans were obtained using both a simple tilted parallel beam (TPB) and a complex projected sine wave (PROJSINE) 3D trajectory (Brzymialkiewicz *et al* 2006) (table 1, figure 3). Absolute activity was again $10 \mu\text{Ci mL}^{-1}$ in the tubes, and activity was continually added to the background to generate the same contrast ratios as for the cylinder experiments. Scan times were initially 10 min and then subsequently adjusted during the acquisition sequences to account for radioactive decay. The initial count rate was ~480 cps for the 10:1 contrast ratio, which was increased compared to the cylinder experiments due to the added volume in the breast background.

During data acquisition, the expected emission counts from the tubes was kept constant for each contrast ratio by adjusting the scan times for radioactive decay, yielding constant noise quality from the tubes in reconstructed images. The continuously added activity to the background, however, results in a greater number of overall acquired counts, and therefore increases the emitted events from the background overall, especially for the cold tube:background contrast ratios (0.4:1, 0.2:1, 0.1:1, which correspond to 1:2.5, 1:5 and 1:10, respectively), therefore yielding improved noise quality for the background. To account for this in the observer study, a separate set of count-normalized 'high noise' images was generated by randomly down-sampling the breast projection data such that the background noise quality was approximately uniform for each concentration ratio. The total targeted counts in each down-sampled projection image were randomly determined using a Poisson distribution with a mean based on the number of counts in each corresponding 10:1 concentration ratio projection image. Two down-sampled projection data sets were created for each concentration ratio to account for sample variation in the count skimming process.

An iterative, ray-driven implemented ordered subsets expectation maximization (OSEM) algorithm, with eight subsets and five iterations was used to accurately model the 3D acquisitions and reconstruct the data. Calculated attenuation correction was applied, assuming a uniform emission volume. A 3D Hann smoothing filter with a cutoff of 0.7 times the Nyquist frequency was then applied to the reconstructed data. For the phantom breast images, the reconstructed volume was rotated in 3D slightly so that the tubes were vertical in the sagittal plane and therefore multiple slices could be combined to enhance image

quality. Single planar images were generated for each experiment set by summing planes where rods were present (fourteen 2.5 mm thick slices).

The resulting summed slice and contiguous 14-slice SPECT image sets were evaluated in random order, in a controlled environment by five independent observers (four medical physicists and one nuclear medicine radiologist). The observers were tasked with identifying the smallest distinguishable rods (details) for all concentrations, noise levels, experimental setups and acquisitions. In this signal-known-exactly (SKE) study, the observers were shown the location and pattern of the rods prior to viewing the study. Three training images were shown prior to beginning the study. One challenge for the reader was to determine whether the randomized images had 'hot' (10:1, 5:1, or 2.5:1 ratios) or 'cold' contrast (e.g. 0.4:1, 0.2:1 and 0.1:1 ratios). Ambient lighting remained low and the observers could not adjust the window and leveling during the study. Summed images were read three times in randomized presentation order to assess intra-observer variability. For each test, results were averaged over the multiple readings.

Intra-observer variability was examined by computing the repeatability coefficient (RC) and intraclass correlation coefficient based on the one-way ANOVA model (ICC1). The RC provides the difference expected for 95% of the contrasts between any two replicated readings. Inter-observer variability was examined by the intraclass correlation coefficient from the two-way ANOVA model with interaction (ICC3 from Barnhart *et al* 2007). The ICC3 value describes the inter-observer agreement between the five observers based on their single readings and it takes intra-observer variability into account. The resulting data were examined by plotting contrast versus the average results by experimental setups. If the reader could not see any rod at all, a value of 7.0 mm was assigned for analyses below. Depending on the best fit to the averaged data (over the multiple readings) from the five observers, linear or quadratic regression was used to model the data from the five observers. The linear lines or quadratic curves were compared between experimental setups by testing the equality of the coefficients from the linear or quadratic functions. Hot and cold contrast curves were analyzed separately. All statistical tests were computed using SAS (version 9.2) statistical analysis software, and a *p*-value less than or equal to 0.05 is considered as statistically significant.

3. Results and discussion

Contrast-detail analysis has known limitations as it is, by nature, subjective and can be susceptible to observer bias. The SKE setup represents the best-case scenario where visualization of the rods should be obvious to the casual observer. Since nuclear images are routinely filtered and smoothed in the clinic, a modest amount of smoothing was used on all image sets. This study did not account for false positives that may be expected to increase in high noise SPECT images. Readers did not give confidence levels for their decisions as in a full ROC observer study. The following results provide a quantitative measure of observer variability in defining the detection limits, comparing varying orbit techniques possible with a fully 3D acquisition-positioning system as well as hot versus cold imaging for a dedicated breast imaging system.

3.1. Summed slice images

Reconstructed images of low noise infinite:1 and 1:10 rods illustrate the best case examples of visualization of rods for these experiments (figure 4). Due to the vertical geometry and axial symmetry of the phantom, the vertical axis of rotation scan optimizes distance-dependent spatial resolution for the rods at all acquisition angles. The increased background volume of the breast phantom creates an additional and non-uniformly varying scatter medium about the rods, visually apparent in the reconstructed images (figure 5).

Contrast-detail observer study results demonstrate an expected trend that finer resolution is observed for higher contrasts (figure 6). For the phantom in the cylinder using the VAOR trajectory, the mean observed detail ranges from ~2.8 mm in the infinite:1 case, which degrades to ~5.2 mm for the 2.5:1 setup. The inverse then occurs from ~3.8 mm for the 0.4:1 case improving to ~3.0 mm for the 0.1:1 concentration ratio. The mean detectable rod size over all the breast acquisitions was ~3.5 mm at the 10:1 concentration ratio, which degraded to ~5.5 mm at the 2.5:1 concentration ratio. For cold contrasts, the average rod detected was ~4.5 mm at 0.4:1, improving to ~3.5 mm. Hot and cold contrasts (e.g. 5:1 versus 0.2:1) generally agreed within ± 0.75 mm from each other, except for the lowest contrast ratios where 0.2:1 was significantly better than 2.5:1. The mean system detection limits observed in this study are approximately in line with the calculated and previously measured spatial resolution for the SPECT system (Brzymialkiewicz *et al* 2005, Madhav *et al* 2009). The system in-plane spatial resolution is determined heavily by the distance-dependent collimator resolution, as well as the intrinsic resolution of the camera, in this case the size of the individual 2.5 mm CZT detector elements.

Comparative test results for statistical significance are summarized in table 2. For cold rod imaging, VAOR exhibited the best results of the three acquisition trajectories. For hot contrasts, TPB outperformed both VAOR and PROJSINE. Observer performance for the contrast-detail phantom in the non-uniformly shaped breast filled was overall close to that of the rod phantom in the uniform cylinder. These studies may have benefitted from the placement of the contrast-detail phantom near to the nipple, thus decreasing the camera ROR and improving resolution, in contrast to placement near the chest wall. A larger breast having increased scatter and attenuation, and a necessarily larger imaging ROR would likely degrade the images and the observer results more.

Intra- and inter-observer results are summarized in table 3, with the average and range of RC and ICC1 values over the five observers for intra-observer variability, and ICC3 values for inter-observer variability. If a repeatability coefficient is 0.7 mm, you expect to see a -0.7 to 0.7 mm difference between any two replicates for 95% of the time. For ICC values, 1.0 indicates perfect agreement and, generally, an ICC value of 0.8 or greater is considered as very good agreement. VAOR scans in the cylinder had the average lowest repeatability coefficient as might be expected. The intra-observer ICC1 values averaged between 0.8 and 0.9 for all of the methods except for the noisy downsampled TPB data. Generally, the complex PROJSINE orbit had the best overall agreement among and within observers for all of these studies.

3.2. Contiguous slice stacks

Contiguous 14-slice stacks of the same data were also included in the study to investigate any perceived differences compared to viewing reduced-noise summed slices. The resulting curves in figure 7 follow the same general trends as the summed slice data, but with an apparent degradation in observed detail of up to ~1 mm for the cold rod images, and a slight improvement for the larger (10:1, inf:1) hot rod contrasts. Notably, for the infinite:1 ratio, the average reader values were 2.2 mm for multi-slice images versus 2.8 mm for the summed images. Readers appeared more able to separate the 2.2 mm rods when scrolling through multiple slices, which may have blurred together when the slices were summed due to inter-slice misalignment in the rods.

Linear regression of the three methods indicates that for contiguous slice stacks, the TPB acquisition method yielded the best results for cold contrasts, but no significant difference between TPB and VAOR scans for positive contrasts (table 2). When testing for statistical significance between summed and stacked data, observers saw significantly smaller objects for summed data for PROJSINE and VAOR, but no significant difference for TPB for cold

contrast ratios. Conversely, for hot contrast ratios there were no significant differences between summed images and slices for PROJSINE and VAOR, but there was a significant difference for TPB.

For all of the breast phantom experimental conditions, the TPB acquisition method demonstrated significantly ($p < 0.05$) superior observer results than for data acquired with PROJSINE. This is likely due to the fact that the TPB acquisition orbit contained many views at a closer proximity to the phantom, while PROJSINE moved farther from the rods due to variations in both polar angle and ROR (table 1). One important caveat is that TPB demonstrated higher intra- and inter-observer variability compared to PROJSINE (table 3). The PROJSINE trajectory has been shown to have more complete sampling throughout the breast volume compared with TPB (Brzymialkiewicz *et al* 2006), and the symmetry of the rods in the phantom should have yielded a consistent trend in observer detection. With the development of automated object contouring on the SPECT system (Cutler *et al* 2008), we would expect better and more uniform overall resolution through the object of interest, which is additionally inferred from the high ICC3 value for the PROJSINE acquisitions (table 2).

3.3. Downsampled images

Downsampled ‘high’ noise data sets suffered from a lower number of total counts, and observer detection performance correspondingly degraded with decreasing contrast and increased noise. Figure 8 compares both contiguous and summed TPB acquisition downsampled data with the original data. For these analyses, if the reader (infrequently) could not see any rod at all, an estimated value of 7.0 mm was assigned. Thus, values located above the dotted 6.2 mm line are estimates, indicating that some of the observers saw no rods at all under those conditions. A value of 8.0 mm was also tested to assess the statistical sensitivity of the assigned value, with no substantial changes to the regression results in table 2. For higher noise data at low contrasts, further experiments with larger-diameter objects are warranted to verify the true lower limit of detectability. The gap between the full data and count-skimmed data is much wider for cold contrasts, because a greater number of counts were skimmed in order to normalize to the 10:1 contrast data. Count normalization additionally results in lower relative signal to noise ratios because in the cold contrasts, counts are spread across the background instead of concentrating in the rods in the positive contrasts. PROJSINE acquisitions (not shown) followed a similar trend. TPB was prone to larger observer variability for the high noise data, demonstrated by the largest repeatability coefficients and ICC values (table 3). Though a definite separation in mean values between original and high noise data sets exists, the overall observer study results are still promising for clinical detection applications where detected count densities are more similar to the high noise data.

4. Conclusions

A high-performance clinical imaging system for dedicated SPECT breast imaging was successfully evaluated with a newly designed and constructed contrast-detail phantom. The average detail in a moderately sized 500 mL breast volume under relatively low noise conditions at 10:1 was ~3.4 mm, degrading to ~5.2 mm with the 2.5:1 concentration ratio. While far from the high resolution of radiographic imaging techniques, the spatial resolution of the system is better than whole body SPECT and PET detectors and comparable to other dedicated nuclear medicine based detectors (Schilling *et al* 2008). Compact CZT-based pixilated cameras are recently commercially available with even smaller pixels than those used on our current camera. Decreasing pixel sizes results in a further drop in sensitivity and increase in noise, which could potentially also affect lesion visibility for low contrast, low count (high noise) lesions. For example, More *et al* found that a larger pixelated (3.2 mm)

NaI camera coupled with a high resolution collimator was better suited for 2D breast imaging than a smaller 1.4 mm pixilated NaI camera (More *et al* 2006); this corroborates our previous similar conclusions with NaI-based pixellated systems for 3D breast imaging (Tornai *et al* 2005a). Hruska *et al* use a high resolution 1.6 mm CZT gamma camera and have increased their overall sensitivity by optimizing (reducing) collimator length as well as adding a second camera head which images closer to the distal lesions of the first head (Hruska *et al* 2008, Weinmann *et al* 2009), so it is still unclear if a smaller pixel would benefit dedicated 3D imaging. If noise could be normalized between 2D and 3D imaging, for example, then one of the main advantages of 3D imaging which is the inherently increased signal contrast due to the removal of overlapping tissues is still obtained. Additional advantages include potential for quantification as well as localization for biopsy targeting.

Overall, the observer study results are promising, demonstrating the potential of detecting sub-centimeter objects of various biological uptake, a critical target for effective functional breast imaging. For improved accuracy and less subjective results, a larger sample set of images is needed to more completely characterize the system. More realistic, 3D spherical lesions should also be evaluated at various contrasts for a variety of breast sizes in a broader observer study, since there are some advantages here using symmetric rods. Summing multiple planes to reduce noise improved detectability by up to ~1 mm smaller object visualization compared to objects in sliced-stacks, except for the highest 'hot' contrasts (>10:1), where multi-slice stacks improved visualization of the smallest rods. Spatial resolution correlated with the radius of rotation of the camera, highlighted for example by the observed finer resolution for TPB versus PROJSINE scans, because TPB contained more views at a closer proximity to the phantom. PROJSINE, however, had the most consistent combined intra- and inter-observer results, making it the best imaging approach for consistent results.

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Figure 1.

Photograph of the prototype dual-modality dedicated breast imaging system. The SPECT system (center) is placed orthogonally to the CT tube (right) and digital flat-panel detector (left). The colored arrows illustrate system motions (azimuthal for SPECT and CT, and polar and ROR for the SPECT subsystem). A custom steel bed supports the patient as her pendant breast is imaged in the common field of view of the systems.

Figure 2.
(Left) 3D rendered design of the contrast-detail phantom frame with rod diameters (mm).
(Middle) Thin-walled PTFE tubing showing two different diameters. (Right) Contrast-detail phantom in water-filled cylinder imaged using a simple circular scan at a close radius of rotation. Food coloring was added to the fluid inside the tubes for better visibility.

Figure 3.

Acquisition orbits, TPB and PROJSINE used for the phantom in breast studies. To the left are visual representations of the 3D orbits, and to the right are polar plots of the camera tilt as a function of an azimuthal angle, where the radius refers to polar tilt, and in-plane angular displacement represents azimuth.

Figure 4.

Reconstructed images (5th iteration, 8 subsets, 14 summed slices, calculated attenuation correction, post-reconstruction Hann smoothing applied at $0.7 \times \text{Nyquist}$ to reduce noise) of the phantom in the cylinder for infinite:1 (left) and 1:10 (right) concentration ratios. Rod sizes are indicated in mm. Note that three additional cold spots are seen on the right (2 (at the arrow tip), 6 and 10 o'clock) due to the 5 mm diameter posts holding the tube alignment layers together.

Figure 5.

Reconstructed summed coronal slice images for rod:background concentration ratios ranging from 10:1 down to 0.1:1 (effectively 1:10) for VAOR scans about the cylinder, and two trajectories about the breast phantom. Downsampled reconstructions for the TPB orbit (only) are shown on the bottom row, where projection count densities were normalized to the 10:1 concentration case.

Figure 6. Contrast-detail plots illustrating mean and standard deviation values of independent observers comparing the rod phantom imaged in a cylinder using a standard VAOR trajectory and in a breast using PROJSINE and TPB acquisition trajectories. The upper limit of the phantom (6.2 mm) is indicated with a dotted line.

Figure 7.

Contrast versus detail plots illustrating mean and standard deviation of independent observers comparing contiguous slice images of the rod phantom imaged in a cylinder with a standard trajectory and in a breast using PROJSINE and TPB trajectories. The upper limit of the phantom (6.2 mm) is indicated with a dotted line.

Figure 8.

Contrast-detail plot for the TPB orbit for contiguous slice and summed slice images, original and 'high noise' downsampled data. The upper limit of the phantom (6.2 mm) is indicated with a dotted line.

Table 1

Parameters used for acquisitions over a 360° azimuthal range (θ).

Orbit	No of projections	Polar tilt, ϕ (range, min:max)	ROR (range, min:max (cm))
VAOR	128	0°	4.3
TPB	128	45°	6.3
PROJSINE	128	15–45°	3.3–7.6

Table 2
Summary of linear/quadratic regression tests for statistical significance ($p > 0.05$ highlighted)

Methods	p value	Summed Slices			Original versus Downsampling			
		TPB versus PROJSINE	VAOR versus PROJSINE	TPB versus VAOR	PROJSINE (summed)	TPB (summed)	PROJSINE (slices)	TPB (slices)
	Cold	0.0232	<0.0001	0.0363	<0.0001	<0.0001	<0.0001	<0.0001
	Hot	<0.0001	<0.0001	0.0074	0.0053	0.0015	0.1754	0.0276
Summed versus Slices								
	p value	TPB versus PROJSINE	VAOR versus PROJSINE	TPB versus VAOR	PROJSINE	TPB	VAOR	
		0.0003	0.5055	0.0035	<0.0001	0.0971	<0.0001	
	Hot	0.0046	0.1418	0.3623	0.7078	0.0125	0.1323	

Table 3

Summary of intra/inter-observer variability results

Method	Intra-observer		Inter-observer ICC3
	Average RC (range) (mm)	Average ICC1 (range)	
PROJSINE	0.89 (0.48–1.59)	0.91 (0.84–0.97)	0.83
TPB	0.77 (0.00–1.33)	0.80 (0.49–1.00)	0.71
VAOR	0.67 (0.00–1.38)	0.86 (0.58–1.00)	0.68
Downsampled PROJSINE	0.93 (0.72–1.48)	0.91 (0.83–0.97)	0.89
Downsampled TPB	1.96 (1.41–2.95)	0.36 (0.18–0.55)	0.40