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Characterizing the MTF in 3D for a Quantized SPECT Camera Having Arbitrary Trajectories

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Abstract

The emergence of application-specific 3D tomographic small animal and dedicated breast imaging systems has stimulated the development of simple methods to quantify the spatial resolution or Modulation Transfer Function (MTF) of the system in three dimensions. Locally determined MTFs, obtained from line source measurements at specific locations, can characterize spatial variations in the system resolution and can help correct for such variations. In this study, a method is described to measure the MTF in 3D for a compact SPECT system that uses a $16 \times 20 \text{ cm}^2$ CZT-based compact gamma camera and 3D positioning gantry capable of moving in different trajectories. Image data are acquired for a novel phantom consisting of three radioactivity-filled capillary tubes, positioned nearly orthogonally to each other. These images provide simultaneous measurements of the local MTF along three dimensions of the reconstructed imaged volume. The usefulness of this approach is shown by characterizing the MTF at different locations in the reconstructed imaged 3D volume using various (1) energy windows; (2) iterative reconstruction parameters including number of iterations, voxel size, and number of projection views; (3) simple and complex 3D orbital trajectories including simple vertical axis of rotation, simple tilt, complex circle-plus-arc, and complex sinusoids projected onto a hemisphere; and (4) object shapes in the camera's field of view. Results indicate that the method using the novel phantom can provide information on spatial resolution effects caused by system design, sampling, energy windows, reconstruction parameters, novel 3D orbital trajectories, and object shapes. Based on these measurements that are useful for dedicated tomographic breast imaging, it was shown that there were small variations in the MTF in 3D for various energy windows and reconstruction parameters. However, complex trajectories that uniformly sample the breast volume of interest were quantitatively shown to have slightly better spatial resolution performance than more simple orbits.

Index Terms

CZT detector; emission imaging; mammotomography; modulation transfer function; single photon computed emission tomography; three-dimensional trajectories

I. Introduction

The Modulation Transfer Function (MTF) can be used to characterize the frequency composition of spatial resolution. For emission imaging, the MTF can be determined via the line spread function (LSF) from images of a line source [1] (i.e. a capillary tube filled with radioactivity) or an edge source (i.e. radioactivity placed behind a lead plate) [2] in air. Measurements are usually taken at various distances from the camera face to determine spatial variations in resolution. MTF variations with respect to object-to-background ratio contribute to characterizing system linearity. Knowledge of these variations in the spatial resolution of an imaging system is important in understanding the reliability of information derived from an image, (i.e. gauging the minimum size of a lesion that could be detected).

In emission tomography, spatial resolution is determined quantitatively by measuring the LSF or point spread function (PSF), and semiquantitatively by assessing the reconstructed image quality of standard test objects such as the 3D disk and frequency-resolution (cold or hot) rod phantoms. In 2D, measurements of the LSF or PSF often assume that radiation detectors have a continuous response across their surface that is invariant within any given plane parallel to the detector's surface. However, this invariance assumption is questionable for recent medium- and small-field-of-view nuclear emission cameras having quantized detector elements, as are assumptions of approximately Gaussian response. Parallel hole or pinhole collimation, e.g. for small-animal imaging, involves significant nonstationarity in gamma camera response. It is also more common to describe the system performance in the frequency domain by obtaining the Fourier Transform of the PSF or LSF. Generally, image filtering is done in frequency space. The noise power spectrum (NPS), detective quantum efficiency (DQE), and other common parameters used in describing system response are also all measured in the frequency domain and require the MTF in their calculations. Thus, there are various reasons for computing the MTF instead of simply the PSF or LSF. Therefore, by looking at the frequency response, measurement of the MTF in 2D or 3D can provide a quantitative insight into the full frequency response of the system regardless of the detector response function.

Previous line-source phantoms used to measure the MTF in 3D have been limited to measuring the spatial resolution one dimension at a time [3]-[5]. Arrays of point sources have also been used to assess spatial resolution in three dimensions at various locations throughout the image volume [6]. However, unlike line sources, 3D arrays of point sources are more difficult to manufacture and position uniformly, and also are more difficult to evaluate under different scatter and scatter-free conditions due to the nature of their support structure. For this work, a novel phantom was constructed to fully evaluate the three-dimensional MTF along three nearly orthogonal axes after a single tomographic acquisition. This phantom allows for the characterization of spatial resolution in three dimensions and at various locations throughout the imaged volume, providing insight into the total amount of blur and its spatial variation. This phantom can be especially useful for tomographic imaging systems capable of fully-3D acquisitions for SPECT or PET, though it could also be modified for use in transmission imaging systems [7].

Over the past few years, our laboratory has developed a dedicated emission mammotomography system [8]-[11]. Several 3D, non-traditional, noncircular trajectories about a pendant breast with various lesion sizes located in different areas have been implemented and assessed [9], [12]. Due to the spatially varying spatial resolution in SPECT and the nonlinear response of iterative ordered-subsets expectation maximization (OSEM) reconstruction, the system blur cannot be fully described by a single LSF or MTF. Instead, some indication of system blur can be provided by measuring the blur at several locations in

the image, for several signal-to-background ratios, and for several sizes and shapes of structures of interest.

The purpose of this study is to describe the novel phantom and to use this phantom to characterize the local MTF in 3D and to identify parameters that can affect the overall spatial resolution of our current SPECT system. Here we focus on measuring the MTF in different locations in the 3D volume using several different energy windows, different reconstruction parameters and acquisition trajectories, and different object shapes. In the present paper, LSF refers to the image of a line source in zero background and at a particular location. MTF refers to the absolute magnitude of the Fourier Transform of a particular LSF (i.e. at a particular location in the image) and thus throughout this paper the MTF will be referred to as the local MTF. In addition, isotropic in this paper is defined in terms of the MTF being uniform throughout the entire volume.

II. Materials & Methods

A. MTF Phantom

In order to quantify the spatial response of any imaging system including detector response, image acquisition procedures, complex 3D trajectories, scatter, and photon attenuation, a novel phantom was constructed (Fig. 1) to measure the local MTF in 3D. This phantom could be used for both emission (SPECT, PET, or even MRI) and transmission (X-ray CT) systems [7], [13]. A line source was chosen from which to obtain the MTF, since a radioactivity-filled capillary tube for emission imaging and solid tungsten wire for transmission imaging are easily interchangeable. Thus, this phantom would allow using a similar acquisition and quantification procedure for most imaging systems. More than one line source was necessary such that the MTF could be measured at different locations in the reconstructed imaged volume after a single tomographic acquisition. Using too many lines sources runs the risk of creating an overlap between the line sources in the reconstructed images, and so three lines were chosen. Additionally the phantom had very low scatter and attenuation characteristics, could be easily assembled, could be placed in any desired position in the camera's field of view, could be inserted into various fluid filled containers, and could provide flexibility to reposition the line sources in order to image anywhere in the imaged volume.

Based on the desired features, acrylic rods (side length of 7.8 cm) are attached together into the shape of a box frame. Three capillary tubes (~ 11 cm long with 0.9 mm inner diameter) filled with 44.4–66.6 MBq of aqueous ^{99m}Tc (140 keV) were placed within the boxed frame such that each tube was held from one vertex to its corresponding opposite end, slightly away from another vertex. On its corresponding opposite end, the tube is attached to a slider piece which gives the flexibility to position the tube anywhere along the length of the side-rod to which it is attached. Once the desired position of the rod was set, the sliding support was fixed with an acrylic screw pressed against the side-rod. Depending on the size of the SPECT system, pixel size of the camera, and its minimum radius of rotation, smaller or larger capillary tubes could be used to encompass the entire field of view. The tubes were visibly separable in their positions such that they did not touch one another, and spanned the entire field of view. Also, they were not placed at exactly orthonormal planes in order to avoid interplay of measured responses from different directions near the center of the field of view.

B. Data Acquisition and Image Reconstruction

Our emission tomography system is composed of a compact $16 \times 20 \text{ cm}^2$ field of view cadmium zinc telluride (CZT) LumaGEM 3200S gamma camera (*Gamma Medica, Inc.*,

Northridge, CA) attached to a hemispherically positioning gantry. The camera uses a 64×80 array of $2.5 \times 2.5 \times 6$ mm³ quantized CZT elements for a total of 5120 pixels. Measured mean energy resolution of the gamma camera at 140 keV is 6.7% FWHM and overall sensitivity is 37.9 cps/MBq [11]. A parallel-hole collimator with hexagonal holes (1.2 mm hole size flat-to-flat (inner diameter), 0.2 mm septa, and 25.4 mm height) is used in these studies. Due to the flexible gantry, the camera can be positioned anywhere in a hemisphere to facilitate acquiring projection data around a pendant uncompressed breast [8], [9], [11], [12].

Projections were collected over a 360° azimuthal rotation around the phantom suspended in air [Figs. 1 and 2(a)] using a vertical axis of rotation (VAOR), comparable to the simple circular orbits used by clinical SPECT systems (about their horizontal axis). Acquisitions were made at a radius of rotation (ROR) that corresponded to the minimum ROR possible when the MTF phantom was placed in different object shaped media (described in the next section). Data was collected in list mode and post-processed to obtain $\pm 4\%$ and $\pm 8\%$ wide symmetric energy windows about the 140 keV photopeak. Total scan time for each acquisition was 10 min, with increased acquisition times for subsequent scans to compensate for radioactive decay. Under these conditions, there were ~28 k events per projection in the $\pm 8\%$ wide energy windowed data.

Images were iteratively reconstructed using 8 subsets of OSEM, and displayed at 1, 3, 5, and 10 iterations in all cases. These reconstructions were on a grid of $80 \times 80 \times 80$ voxels, for two different voxel widths—1.25 mm and 2.50 mm. The reconstructions were performed using the same ray-driven code as described earlier in [9], [10].

Three other trajectories were investigated: (1) tilted parallel beam (TPB) with 45° fixed camera tilt; (2) tilted circle plus (one) arc (CPA) with 45° max camera tilt, and (3) a three-lobed sinusoid projected onto a hemisphere (PROJSINE) with 0° to 45° polar camera tilting range (sinusoidal amplitude). These trajectories have been investigated elsewhere [9], [11], [12], but not for their frequency response characteristics. These trajectories and the direction of polar camera tilt are illustrated in Fig. 2. Specific parameters that were used for all four trajectories are given in Table I. The increased number of projections is due to the use of polar (ϕ) sampling along with azimuthal (θ) sampling, necessitating $\Delta\phi$ steps along with $\Delta\theta$ steps to reach 360° azimuthally.

C. MTF Phantom in Different Object Shapes

The degrading effects of different object shapes on the MTF were also evaluated by placing the line source phantom in two different environments: 2000 mL water-filled glass-walled cylinder (12 cm diameter) and 850 mL water-filled plastic-walled breast-shaped container (nipple-to-chest distance of 11 cm, medial-lateral distance of 17 cm, and superior-inferior distance of 18 cm) (Fig. 3). All four trajectories were investigated except for the cylinder where only VAOR was used. The local MTF was calculated for all trajectories measured with the phantom placed in the cylinder and breast. Data was collected using an $\pm 8\%$ wide symmetric energy window and was post-processed to also obtain $\pm 4\%$ wide symmetric energy windows. Energy window width for dedicated, uncompressed breast imaging has been shown to affect image quality under various acquisition conditions [14]. The $\pm 8\%$ wide energy window, while not exploiting the fine resolution capability of CZT (6.7% intrinsic FWHM), corresponds to that used in standard, clinical nuclear medicine camera systems (15% wide), and is substantially narrower than that of scintillator based compact, quantized detector element gamma cameras (30% wide). Furthermore, early contrast-detail observer studies with this system have shown that energy windows near but above the intrinsic system resolution allow for visualization of small objects [14]. The ROR was determined by placing the breast 1 cm or less away from the camera face at all camera orientations. For all

four trajectories tested, acquisitions were obtained at the same RORs used for the phantom in air (shown in Table I). Scans were also performed over 10 min, with increased acquisition times for subsequent scans to compensate for radioactive decay. Under these conditions, approximately 13 k counts were collected in each $\pm 8\%$ wide energy windowed projection of the MTF phantom measured in the scatter media. Iterative reconstruction was performed using the same parameters for the projections of the phantom suspended in air.

D. Local MTF Determination

The overall local MTF calculation was computed based on an algorithm presented elsewhere [15], and has been successfully applied in the preliminary local MTF measurements of our 3D X-ray computed mammotomography (CmT) system as well [7]. Once the images were reconstructed, the first step was to manually rotate the images such that each tube could be isolated onto a single slice [Fig. 2(c)]. A binary line image of the outline of the tube was acquired by using a Sobel edge detection method [16]. The angle and position of the tube was determined by the Radon transform. Based on the resulting curves in polar coordinate space, the location of the curves' intersection (i.e. maximum intensity) gave an estimate of the angle and spatial position of the line source [17]. Inaccuracy in estimating the true angle of a line can give an unsmooth LSF and degrade the MTF, especially in higher frequency regions. A composite local LSF was generated by reprojecting and resampling a small region of interest around the line source onto a one-dimensional array of sub-pixel bins positioned perpendicular to the calculated angle of the tube. Use of subpixel bins was necessary since the samples were not uniformly distributed along the one-dimensional array [18]. Different subpixel bin sizes (0.1, 0.3, and 0.7) were initially tested to find an optimal bin size. For smaller bin sizes, such as 0.1, high frequency components were exaggerated in the MTF due to the increased noise. At larger bin sizes, such as 0.7, the zero of the MTF occurred below the Nyquist frequency. Therefore, for this work 0.3 subpixel bin size was found to be the optimal size. The samples were thus binned to 1/3 of the original pixel size of 2.5 mm. This result was then smoothed by averaging neighboring pixels within a three pixel moving average window to get the final local LSF (LSF_f)

$$LSF_f(x) = \frac{LSF(x-1) + LSF(x) + LSF(x+1)}{3} \quad (1)$$

where x is the spatial location of the pixel. Smoothing is necessary to reduce the noise and effect of the high-frequency response in the MTF. Zero padding and a Hanning filter were also applied to the local LSF to eliminate other high frequency artifacts.

Corresponding local MTFs were calculated by taking the absolute value of the one-dimensional discrete Fourier Transform of the LSF_f

$$MTF(f) = |\mathcal{F}\{LSF_f(x)\}| \quad (2)$$

where f represents the spatial frequency. The local MTF here describes the magnitude of the component response of the system to the known input. Since an infinitely small line source is not being used, the effects of the finite width of the capillary tube, a , is corrected for by dividing the MTF by $a \text{sinc}(\pi a f)$. This MTF was finally normalized to yield the final MTF. The MTFs along various parts of the tube (e.g. a "local MTF") were calculated in order to evaluate the response at different locations in the sampled volume [Fig. 2(c)].

For each tube, an average MTF, its standard deviation, and upper and lower bounds for the estimated average MTF was calculated from the measured MTF along different segments of

the tube. Using these results, the root mean square difference (RMSD) was calculated by subtracting the upper and lower bounds, squaring the difference, finding the average of all these squared differences, and taking the square root. The measured RMSD values help to evaluate the amount of uniformity in the MTF along different portions of a single line source.

III. Results and Discussion

A. MTF Along Different Line Location

Fig. 4 shows the local MTFs measured in air with the $\pm 8\%$ wide energy window at different locations along each of the nearly orthogonal capillary tubes. Measurements were derived from images collected of the phantom in air using the VAOR orbit. For all three orientations, there was only a small variation in the MTF along the lines, indicating uniform 3D resolution throughout the imaged volume. VAOR orbits are known to be the largest uniformly sampled volumes for both conventional and compact, versatile 3D SPECT cameras [9]. The uniformity of these results is not surprising since the SPECT scan was collected over a 360° azimuthal orbit. Hence, projection “averaging” throughout this reconstructed volume may be expected to yield a nearly uniform response.

B. MTF at Different Energy Window

The MTF was also determined at $\pm 4\%$ and $\pm 8\%$ energy windows to see if different amounts of object scatter would affect the resolution of the reconstructed images. Fig. 5 shows the MTF for one of the capillary tubes in the MTF phantom suspended in the water-filled cylinder and the water-filled breast, with data acquired using VAOR and TPB orbits at both energy windows. For the cylinder, there is no difference in the MTFs measured anywhere along the line.

For the breast phantom, the MTF results are somewhat different. There are differences in the measured MTFs at different locations, which correspond to the thicker posterior breast (top) or narrower anterior breast near the nipple (bottom), but only slight differences in these locations due to energy window width. Furthermore, using a smaller energy window with incomplete sampling and the TPB orbit also had a very slight effect on the local MTF.

C. MTF With Various Reconstruction Parameters

In OSEM, the projection data is grouped into subsets and the EM algorithm is applied to each subset to get an estimate of the image. In principle, more iterations are performed until the algorithm has converged to a true (or near) optimum value. However this can yield images of poor quality. Previous studies for emission mammotomography imaging breast lesions have shown that there is a trade-off between the rate of contrast improvement and SNR degradation in the reconstructed images [8], [9], [12]. This contrast-SNR roll-over is similar to the minimum of a bias-variance metric. For SPECT mammotomography imaging this roll-off is often best between 2 and 3 iterations.

In a similar way, the measured local system MTF appears to have stopped changing after only a few iterations. Fig. 6 shows the MTF at 1st, 3rd, 5th, and 10th iterations along a single dimension. The slow-down in MTF improvement is probably similar to the slow-down in contrast improvement with increasing iterations. At the 5th and 10th iteration, it appears that there is no change in the MTF beyond the 3rd iteration. The results are similar (not shown) for the other portions of each of the three capillary sources for a uniform orbit such as VAOR.

Using smaller voxel sizes and fewer projection images during iterative reconstruction can also have an effect on the measured resolution (Fig. 7). As voxel size decreases and number of projections increase, the MTF expectedly improves and remains consistent (less anisotropic), especially under nonideal conditions (i.e. water-filled cylinder) (Fig. 7, BOTTOM). From these plots, it can be inferred that when acquiring fewer projections under non-ideal conditions, the MTF varies significantly due to incompleteness of sampling the scatter distribution and can introduce artifacts in the reconstructed images. However, very small differences are seen in the MTF for conditions without scatter (Fig. 7, TOP). The MTF obtained at 128 projections in the water-filled cylinder appears to be better than in air. This could be due to the increased attenuation and noise in the scatter images which made it more difficult to localize the signal.

D. MTF With Various 3D Trajectories

Other than VAOR, three different trajectories, TPB, CPA, and PROJSINE, were also evaluated by measuring the MTF at three different locations along the capillary tube in air (Fig. 8). RMSD results in Table II indicate small but similar variations in the MTF for VAOR and PROJSINE. The importance of this is that the complex trajectory will likely be necessary for clinical breast imaging [8], [9], while VAOR provides a completely sampled region with the largest volume. The TPB and CPA trajectories yielded a worse MTF in the top portion, but had a better MTF at the bottom portion due to the relative position of the camera to the line sources (Fig. 8). This is most likely due to a combination of incomplete sampling, attenuation, and distance related degradations associated with SPECT imaging.

Fig. 9 shows the difference in spatial resolution among each of the four orbits for the top segment of the capillary tube. Previous studies have shown that insufficient sampling can cause the reconstructed slices to be distorted [9], [12]. For the TPB orbit, in contrast to the bottom segment of the capillary source, the top segment of the capillary tube [shown in schematic in Fig. 2(a)] is never close to the face of the SPECT camera, causing a difference in the measured spatial response.

E. MTF Within Different Object Shapes

Using a VAOR orbit, the MTF was determined for the line phantom suspended in (1) air; (2) a water-filled cylinder; and (3) a water-filled breast phantom using the same, fixed ROR for all three cases. Images reconstructed with 5 iterations and 5 subsets of an OSEM algorithm produced minimal differences between the MTF obtained in air and in the cylinder (Figs. 4 and 7), but showed a slight difference in comparison to the MTF obtained with the breast phantom (Fig. 10) mostly likely due to the differences in object shape (i.e. breast phantom versus a cylinder) and a slight difference in the measured location relative to the measurement in air or with the cylinder. RMSD values for capillary tube B acquired in air, cylinder, and breast are 0.044, 0.048, and 0.064, respectively. This slight degradation in resolution seen with VAOR in a breast phantom illustrates the necessity of using novel trajectories that use increased camera tilt angles to improve sampling and also achieve a closer ROR. As shown in Figs. 10, 11, and Table III, there is slight variation in the MTFs with VAOR, TPB and CPA trajectories, consistent with earlier results using these trajectories. There is considerably less variation with PROJSINE. Comparisons of all four trajectories are plotted together for the top and bottom segment of the capillary tube (Fig. 12).

IV. Conclusion

In this work, we have developed a novel phantom for use in measuring the local MTF in 3D to evaluate versatile, novel trajectory emission tomographic systems and provide an insight

into the degrading effects caused by different orbital acquisitions, different shaped geometries, and the reconstruction process. This initial study demonstrated that energy window has a secondary effect on resolution, in contrast to incomplete sampling. The relatively small number of OSEM iterations also had little effect on the MTF, and that more projections are necessary in non-ideal conditions (i.e. with attenuation media) to yield an improved and more isotropic MTF. Also, it was shown that unlike for simple circular (VAOR) and more complex (PROJSINE) trajectories, asymmetric acquisition trajectories (e.g. TPB and CPA) have more variations in the MTF between different locations in 3D space. Differences between the curves are caused both by trajectories having fixed polar tilt and subsequently incomplete sampling. This implies that a complex trajectory like PROJSINE more uniformly samples the breast volume of interest and may be necessary for clinical breast imaging. Having flexible 3D positioning about the breast yielded minimal RMSD differences, which is important for high resolution molecular emission imaging.

This study also demonstrates the benefit of using a local 3D MTF phantom to evaluate a tomographic imaging system. With such a phantom, the effect of attenuation, scatter, energy windows, reconstruction process, and 3D trajectories can be easily evaluated to get a better understanding on how these different parameters can affect image quality.

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References

1. Nusynowitz ML, Benedetto AR. Simplified method for determining the modulation transfer function for the scintillation camera. *J Nucl Med* 1975;16:1200–1203. [PubMed: 1194976]
2. Väyrynen T, Pitkänen U, Kiviniitty K. Methods for measuring the modulation transfer function of gamma camera systems. *Eur J Nucl Med* 1980;5:19–22. [PubMed: 7379801]
3. Coleman M, King MA, Glick SJ, Knesaurek K, Penney BC. Investigation of the stationarity of the modular transfer function and the scatter fraction in conjugate view SPECT restoration filtering. *IEEE Trans Nucl Sci* 1989;36:969–972.
4. Glick SJ, King MA, Knesaurek K, Burbank K. An investigation of the stationarity of the 3D modulation transfer function of SPECT. *IEEE Trans Med Imag* 1989;36:973–977.
5. Glick SJ, Hawkins WG, King MA, Penney BC, Soares EJ, Byrne CL. The effect of intrinsic attenuation correction methods on the stationarity of the 3-D modulation transfer function of SPECT. *Med Phys* 1992;19:1105–1112. [PubMed: 1518474]
6. Furenlid LR, Wilson DW, Chen Y, Kim H, Pietraski PJ, Crawford MJ, Barrett HH. FastSPECT II: A second-generation high-resolution dynamic SPECT imager. *IEEE Trans Med Imag* 2004;51:631–635.
7. Madhav, P.; McKinley, RL.; Samei, E.; Bowsher, JE.; Tornai, MP. A novel method to characterize the MTF in 3D for computed mammotomography. *Proc. SPIE Med. Imag. Conf.*; Feb. 11–17, 2006;
8. Tornai MP, Bowsher JE, Archer CN, Peter J, Jaszczak RJ, MacDonald LR, Patt BE, Iwanczyk JS. A 3D gantry single photon emission tomograph with hemispherical coverage for dedicated breast imaging. *Nucl Instrum Meth Phys Res A* 2003;497:157–167.
9. Archer CN, Tornai MP, Bowsher JE, Metzler SD, Pieper BC, Jaszczak RJ. Implementation and initial characterization of acquisition orbits with a dedicated emission mammotomograph. *IEEE Trans Nucl Sci* 2003;50:413–420.
10. Brzymialkiewicz CN, Tornai MP, McKinley RL, Cutler SJ, Bowsher JE. Performance for dedicated emission mammotomography for various breast shapes and sizes. *Phys Med Biol* 2006;51:5051–5064. [PubMed: 16985287]

11. Brzymialkiewicz CN, Tornai MP, McKinley RL, Bowsher JE. Evaluation of fully 3D emission mammotomography with a compact cadmium zinc telluride detector. *IEEE Trans Med Imag* 2005;24:868–877.
12. Brzymialkiewicz, CN.; Tornai, MP.; McKinley, RL.; Bowsher, JE. 3D data acquisition sampling strategies for dedicated emission mammotomography for various breast sizes. *Proc. IEEE Nuclear Science Symp. Medical Imaging Conf.*; Oct. 16–22, 2004; p. 2596-2600.
13. Madhav, P.; Brzymialkiewicz, CN.; Cutler, SJ.; Bowsher, JE.; Tornai, MP. Characterizing the MTF in 3D for a quantized SPECT camera having arbitrary trajectories using a novel phantom. *Proc. IEEE Nuclear Science Symp. Medical Imaging Conf.*; Oct. 23–29, 2005; p. 1722-1726.
14. Cutler, SJ.; Brzymialkiewicz, CN.; Tornai, MP. Investigating the effects of energy resolution in dedicated emission mammotomography. *Proc. IEEE Nuclear Science Symp. Medical Imaging Conf.*; Oct. 23–29, 2005; p. 2537-2541.
15. Samei E, Flynn MJ. A method for measuring the presampled MTF of digital radiographic systems using an edge test device. *Med Phys* 1998;25:102–113. [PubMed: 9472832]
16. Lim, JS. *Two-Dimensional Signal and Image Processing*. Englewood Cliffs, NJ: Prentice-Hall; 1990.
17. Gonzalez, RC. *Digital Image Processing*. 2. Reading, MA: Addison-Wesley; 1987.
18. Fujita H, Tsai D, Itoh T, Doi K, Morishita J, Ueda K, Ohtsuka A. A simple method for determining the modulation transfer function in digital radiography. *IEEE Trans Med Imag* 1992;11:34–39.

Fig. 1.

(Left) Photograph of the acrylic phantom frame with three brass rods (brass rods were used in place of capillaries for the photo to clearly display the position of each tube). (Right) Photograph of the phantom suspended in front of the compact SPECT camera in air during a 360° acquisition.

Fig. 2.

(a) 3D setup of the phantom for all four trajectories. The SPECT camera (shown as a gray box) is facing left towards the phantom, placed at the center of the field-of-view. The phantom consists of three radioactivity-filled capillary tubes (labeled A,B,C). The shown placement of the tubes nearly matches the photograph of the phantom suspended in front of the SPECT camera in Fig. 1. The direction of increased polar camera tilt, ϕ , is also shown relative to the horizon. (b) Next to each 3D schematic is the corresponding polar plot (i.e. camera tilt) of the camera trajectory (plotted as a radius with $+90^\circ$ at the center and -30° at the edge) versus azimuth angle (plotted around the circle from 0° to 360°). (c) Reconstructed slice (5th iteration, 8 subsets, $80 \times 80 \times 80$ grid size, 2.5 mm) for one isolated capillary tube located in air acquired with a TPB orbit. “Top,” “Middle,” and “Bottom” of the capillary tube are defined for subsequently presented results, and span 11 cm along the length of the line source.

Fig. 3. Photographs of the MTF phantom contained inside (Left) the 2000 mL cylinder and (Right) 850 mL breast phantom shell. (Bottom) Phantom was immersed in the water to the extent possible to quantify the effect of different object shapes along with various acquisition trajectories on the MTF, and (Top) is also shown in the phantoms without water present.

Fig. 4.

Results for tube A (top), B (middle), and C (bottom) show little variation in MTFs at different portions along the capillary line source for a 360° SPECT acquisition. MTF measurements were obtained from measurements taken of the phantom in air using a simple VAOR orbit. Nyquist frequency is at 0.2 mm^{-1} .

Fig. 5. MTF of the top and bottom segment of the capillary source B placed in a water-filled cylinder (Left) and a water-filled breast shape phantom acquired using VAOR (Middle) and TPB (Right) orbits at $\pm 4\%$ and $\pm 8\%$ energy windows. Tighter energy windows appear to have a slight improvement in resolution especially at the TPB case.

Fig. 6. MTF at several iterations of the phantom in air using VAOR. MTFs overlap for iterations 3, 5 and 10 (shown by arrow). MTF was obtained using the top portion of capillary A.

Fig. 7.

(Top row) MTF obtained of the phantom in air at a reconstructed voxel size of (Left column) 1.25 mm ($160 \times 160 \times 160$ grid size) and (Right column) 2.5 mm ($80 \times 80 \times 80$ grid size). Plots indicate very slight difference between the different numbers of indicated projections. (Bottom row) Unlike in air, the MTF obtained from the phantom in the water-filled cylinder, for both voxelizations, has a greater decrease and more anisotropy in MTF for the 64 projection data set. MTFs shown here were obtained using capillary tube B and VAOR orbital acquisition (see Fig. 2).

Fig. 8.

Using TPB (top), CPA (middle), and PROJSINE (bottom) trajectories, the MTFs were obtained from the top, middle, and bottom segment of the reconstructed line source B. The MTFs for TPB and CPA have more variation among different segments along the tube compared to VAOR (Fig. 4, middle) and PROJSINE. For the bottom line segment, both TPB and CPA actually give a higher MTF than VAOR and PROJSINE due to the bottom segment being closest to the camera throughout the trajectory, as seen in the schematic in Fig. 2(a).

Fig. 9.

MTFs were obtained from the top segment of the reconstructed line source B for each of the three orbits. TPB produces slightly worse MTFs than VAOR and PROJSINE since the top segment is farther away from the camera for those trajectories.

Fig. 10.

MTF obtained from the line phantom (capillary tube B) placed in the breast using the VAOR orbit. Unlike MTF curves obtained in air and a cylinder, there is more variation in MTF in a breast phantom.

Fig. 11. MTF obtained from the phantom (top and bottom segment of capillary tube B) placed in the breast using TPB (top), CPA (middle), and PROJSINE (bottom) trajectories. Results are very similar to results measured in air.

Fig. 12. MTF of the top (left) and bottom (right) segment of capillary source B placed in the breast for all four trajectories.

TABLE I

Trajectory Parameters Used for Each Acquisition Over a 360° Azimuthal Range

Trajectories	Acronym	Number of Projs	Polar Tilt Range (min-max degrees)	ROR Range (min-max cm)
Vertical Axis of Rotation	VAOR	128	0°	8.74
Tilted Parallel Beam	TPB	128	45°	4.05
Circle Plus Arc	CPA	139	0°-45°	4.05-6.48
Three-Lobed Sinusoid Projected onto a Hemisphere	PROJSINE	150	15°-45°	4.05-8.74

TABLE II

RMSD Values for Each Capillary Tube Acquired Using Various Trajectories

Trajectories	Tube A	Tube B	Tube C
VAOR	0.049	0.044	0.039
TPB	0.145	0.123	0.042
CPA	0.103	0.093	0.091
PROJSINE	0.041	0.032	0.040

TABLE III

RMSD Values for Each Capillary Tube in Breast Acquired Using Various Trajectories

Trajectories	Tube A	Tube B	Tube C
VAOR	0.056	0.064	0.086
TPB	0.103	0.092	0.122
CPA	0.092	0.119	0.096
PROJSINE	0.050	0.052	0.046