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Design Optimization of a TOF, Breast PET Scanner

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Abstract

A dedicated breast positron emission tomography (PET) scanner with limited angle geometry can provide flexibility in detector placement around the patient as well as the ability to combine it with other imaging modalities. A primary challenge of a stationary limited angle scanner is the reduced image quality due to artifacts present in the reconstructed image leading to a loss in quantitative information. Previously it has been shown that using time-of-flight (TOF) information in image reconstruction can help reduce these image artifacts arising due to missing angular projections. Our goal in this work is to optimize the TOF, breast scanner design by performing studies for estimating image uniformity and lesion activity uptake as a function of system timing resolution, scanner angular coverage and shape. Our results show that (i) $1.5 \times 1.5 \times 15 \text{ mm}^3$ lutetium oxy-orthosilicate (LSO) crystals provide a high spatial resolution and system sensitivity relative to clinical scanners, (ii) 2/3 angular coverage scanner design with TOF timing resolution less than 600 ps is appropriate for providing a tomographic image with fewer artifacts and good lesion uptake estimation relative to other partial ring designs studied in this work, (iii) a flat scanner design with 2/3 angular coverage is affected more by larger parallax error than a curved scanner geometry with the same angular coverage, but provides more uniform lesion contrast estimate over the imaging field-of-view (FOV), (iv) 2/3 angular coverage, flat, 300 ps TOF scanner design (for short, practical scan times of ~5 mins per breast) provides similar precision of contrast recovery coefficient (CRC) values to a full curved, non-TOF scanner, and (v) employing depth-of-interaction (DOI) measuring detector and/or implementing resolution modeling (RM) in image reconstruction lead to improved and more uniform spatial resolution and lesion contrast over the whole FOV.

Index Terms

Breast scanner; PET; time-of-flight

I. Introduction

A dedicated breast positron emission tomography (PET) scanner can be used for accurately characterizing and monitoring response of early stage breast tumor with high spatial

resolution and sensitivity relative to a clinical, whole-body PET scanner. In recent years [1]–[8], dedicated breast PET devices have been developed aiming to improve spatial resolution and sensitivity over clinical, whole-body PET scanner for breast imaging. Commercially an FDA approved, dedicated device (PEM Flex Solo II, Naviscan PET System) is currently available, which performs focal-plane tomography for image reconstruction. Geometric flexibility in terms of detector placement around the breast and axilla, the variability of the detector separation for different breast sizes as well as the potential to combine with other imaging modalities, provides a compelling rationale to develop a dedicated breast scanner with a limited angular geometry. A major challenge of the limited angle scanner design is the impact on image quality due to incomplete angular coverage, producing artifactual tomographic images with limited quantification capability. In order to achieve improved high quality tomographic images, detector rotation can be considered. But, incorporating detector rotation makes the scanner design more complex and limits the system flexibility as well. In past work [9], we have shown that using time-of-flight (TOF) information in image reconstruction helps lead to fewer image artifacts with good lesion uptake estimation in the limited angle scanner geometry.

Our goal in this work is to perform design optimization for a TOF, breast PET scanner design by systematically evaluating the reconstructed images using fundamental metrics such as the mean-squared-error (MSE) to quantify the integrity of the image. In contrast to our previous study our lesion contrast recovery coefficient (CRC) evaluation investigates the impact of parallax error, limited angular coverage, and a depth-of-interaction (DOI) measurement capability by using lesions placed at multiple positions within the imaging field-of-view (FOV). In addition, in the current work we extend our system design evaluation to dual flat detectors, which have a practical advantage over curved detectors in terms of mechanical assembly. Finally, we also investigate the impact of incorporating point-spread-function (PSF) modeling in the image reconstruction algorithm for resolution recovery in order to maintain the high spatial resolution throughout the FOV of a small diameter scanner using long crystals for high sensitivity.

II. Methods

A. Scanner Design and Simulation Setup

System design simulations were performed using the GEANT4 Application in Tomographic Emission (GATE) Monte Carlo simulations package [10] in order to understand the benefit of TOF in the reconstruction of limited angle PET data. In our simulations four different scanner geometries were modeled as depicted in Fig 1. For curved scanner geometry three PET ring configurations with a ring diameter of 15 cm and an axial FOV of 15 cm were evaluated: a full curved scanner and two partial curved scanners which were the 2/3 ring setup with a 120 degree in-plane angular coverage and the 1/2 ring setup with a 90 degree in-plane angular coverage. For flat panel scanner configuration two flat panel modules are simulated with the same angular coverage as the 2/3 curved scanner. The detector separation and axial length were fixed at 15 cm. The 2/3 flat panel scanner geometry is interesting due to its potentially easier mechanical assembly and practicality if breast compression is needed as well as the capability of imaging the chest wall at close proximity. TOF information was modeled with timing resolutions of 150, 300, 450 and 600 ps as well as non-TOF. We chose 15 mm long lutetium oxy-orthosilicate (LSO) crystals since this crystal length together with the simulated system geometry for a full ring design gives a system coincidence sensitivity of 7.2% that is similar to the sensitivity of the current generation of clinical whole-body scanners (the sensitivities of 2/3 curved, 1/2 curved, and 2/3 flat are 3.2, 1.8, and 3.2%, respectively). We also chose a crystal cross-section of $1.5 \times 1.5 \text{ mm}^2$ to achieve good spatial resolution needed for breast imaging without significantly increasing the detector cost by using very small and long crystals (crystal pitch was $1.56 \times 1.56 \text{ mm}^2$). Also, smaller crystal

cross-section has been shown to have an adverse effect on the detector timing resolution [11], which will reduce the TOF imaging capability if such a breast PET scanner was developed. For DOI measurement capability a 2-level detector with the first crystal layer being 6 mm thick and the second crystal layer being 9 mm thick was simulated for all curved scanners as well as the dual flat panel scanner. The choice of two layer thicknesses was based on the fact that we have similar detection efficiency in the two layers.

B. Simulation Studies

For characterization of reconstructed system spatial resolution we simulated a point source emitting back-to-back 511 keV photons in air as well as in a warm water background. The point source was placed at transverse radial positions of $r = 1, 3, \text{ and } 5$ cm and spatial resolution studies were limited to full ring, non-TOF scanner geometries. For image uniformity as well as contrast evaluation ten independent lesion phantom data sets were simulated. We simulated a 14 cm diameter and 14 cm long water cylinder containing 25 of 5 mm diameter spheres at multiple radial positions (axially centered) with two different activity uptake ratios relative to the background: 4:1 and 8:1 as depicted in Fig. 2. We also simulated ten independent, but equivalent uniform water phantom data sets separately with the same cylinder size as the lesion phantom. Background activity concentration of 361 Bq/cc and a scan time of 40 minutes were simulated. The background activity concentration was chosen based on the reported average ^{18}F -FDG radiotracer concentration in normal breast tissue [12]. The choice of 40 min scan time was made in order to collect high counts for a consistent evaluation of our scanner design without being impacted by statistical effects in the reconstructed image. Unless otherwise stated differently, all our results are presented for data acquired for the 40 min scan time.

C. Image Reconstruction and Parameters

For point source in air simulations, 3D-FRP algorithm [13], which is a direct 3D Fourier reconstruction method with Fourier reprojection, was used for image reconstruction. All other data were reconstructed using an iterative list-mode ordered-subsets expectation-maximization (OSEM) algorithm (33 subsets and up to 15 iterations) using a Gaussian TOF kernel [14] and modified Kaiser-Bessel ('blob') basis functions [15] optimized for the expected spatial resolution of around 1.5 mm. All data corrections (attenuation, scatter, randoms, normalization) were built into the system model. Attenuation map was generated analytically for a cylindrical phantom with the same dimensions as those of the simulated water phantom with the attenuation coefficient set at 0.095 cm^{-1} for 0.511 MeV. In this work we reconstructed only the true coincidence events and so no scatter or randoms corrections were needed. The geometric sensitivity in the FOV is calculated directly during the list-mode reconstruction. Also, for each scanner design a separate uniform cylinder normalization simulation was performed where data were acquired for a high number of coincidence events. The acquired data were used to generate a normalization sinogram in a manner analogous to the techniques used in clinical systems, which was then used in the list-mode reconstruction algorithm.

The iterative OSEM algorithm has also been modified to use a spatially variant point spread function (PSF) for resolution modeling (RM) in a full curved scanner geometry. In this study the PSFs were generated from simulations of point sources in air at radial distances $r = 1, 2, 3, 4, 5, \text{ and } 6$ cm. The resolution model was determined by fitting the radial profile through the point for a LOR at different positions in sinogram space with a single Gaussian function leading to a symmetric but a good fit for the PSF. The simulated PSFs are, however, slightly asymmetric, but the topic of more accurate PSF modeling is beyond the scope of this paper. For reconstructions with resolution modeling the basis functions were

modeled as delta functions convolved with the PSF at the radial position of each delta function.

D. Data Analysis

Reconstructed system spatial resolution for point source in air was measured by drawing profiles in images to calculate full width at half maximum (FWHM) using the standard NEMA NU 4-2008 procedure [16]. For point source in warm background profiles were drawn according to the NEMA NU 4-2008 protocol followed by a Gaussian fit with an offset due to a warm background. Spatial resolution was defined by the FWHM of the Gaussian fit.

For image uniformity evaluation, we investigated the impact of both systematic and statistical properties of TOF reconstruction due to limited angular coverage on the reconstructed images of the lesion phantom. A scalar metric, MSE of the reconstructed lesion phantom f was calculated for a central transverse image slice as

$$MSE(f) = \frac{1}{N} \sum_{i=1}^N \left(\frac{1}{L} \sum_{s=1}^L \left(\frac{f_i^s - g_i}{g_i} \right)^2 \right) \\ = \frac{1}{N} \sum_{i=1}^N \left(\frac{1}{L} \sum_{s=1}^L \left(\frac{f_i^s - \bar{f}_i}{g_i} \right)^2 \right) + \frac{1}{N} \sum_{i=1}^N \left(\frac{1}{L} \sum_{s=1}^L \left(\frac{g_i - \bar{f}_i}{g_i} \right)^2 \right) \quad (1)$$

where f_i^s is the number of counts at the i -th voxel of the reconstructed image from the s -th data copy of $L = 10$ independent data samples, $\bar{f}_i$ is the average number of counts at the i -th voxel of the reconstructed image over 10 independent data samples, g_i is the simulated activity in the i -th voxel of the lesion phantom, and N is the total number of voxels covering the interior of the phantom in the central transverse image slice. The MSE is first calculated as the mean-squared-error for each image voxel where the simulated activity (g_i) was employed as the normalization factor, and then averaged over the entire object within the central transverse image slice. The normalization with the simulated activity was employed to obtain a relative measure of MSE over the entire object. For each image voxel, the MSE in the above equation can be rearranged as the variance (the first term) plus the bias-squared (the second term) of the reconstructed estimator. While the variance displays the noise structure due to statistical fluctuations (estimated through the ten independent data samples), the minimal bias-squared value (average over the whole object) determines the convergence of the reconstruction estimator expressing how close image voxel estimate is to its true value. The bias-squared value includes systematic effects arising due to incomplete sampling as well as parallax error in the detector.

Lesion CRC was used to estimate the quantitative accuracy of the reconstructed images. ROIs of the same size as the sphere diameter (5 mm) were drawn over the hot spheres in the lesion phantom images to measure the mean counts, C_H . Background ROIs were drawn in the uniform cylinder image at the same location as the sphere. The background ROI diameter was set at 10 mm to obtain a more reliable estimate of the mean background counts, C_B . Using NEMA NU 4-2008 protocol the CRC is calculated as

$$CRC = \frac{C_H/C_B - 1}{R - 1} \quad (2)$$

where R is the simulated lesion uptake ratio with respect to background. CRC values were calculated for each lesion as shown in Fig. 2 and averaged over the 10 independent data sets in order to obtain a mean CRC value per lesion.

III. Results

A. Reconstructed System Spatial Resolution

In Fig. 3, we show the results of spatial resolution (FWHM) measurements of point source in air and in a warm background at various radial positions for $1.5 \times 1.5 \times 15 \text{ mm}^3$ LSO crystal in a full curved geometry. The image of point source in air was reconstructed using the analytic, 3D-FRP reconstruction method while the iterative OSEM (non-TOF) was used for reconstructing the image of point source in the warm background. The OSEM results are after 3 iterations, which we show later to be the point at which the bias-squared values in lesion phantoms reaches a minimum. The spatial resolution is an average over the transverse and radial FWHM values. Since we chose a relatively long crystal (15 mm) with respect to the scanner diameter there is a loss in spatial resolution for large radial positions due to parallax error.

B. Impact of TOF on image uniformity in limited angle tomography

In Fig. 4, we show the central slices for the reconstructed lesion phantom images for the four scanner geometries, as described in Fig. 1, as a function of the timing resolution. As the angular coverage is reduced non-TOF partial curved scanners as well as 2/3 flat scanners have image distortions, artifacts and non-uniformity. The two bright vertical strips visible in both the 2/3 and 1/2 curved scanner non-TOF reconstructions are due to the acceptance of LORs within the same detector at its two edge. Improved timing resolution in TOF images mitigates some of these distortions and artifacts as the scanner angular coverage is reduced. Fig. 5 shows the bias-squared and the MSE of both TOF and non-TOF reconstructed images for curved as well as flat scanners with various angular coverages as a function of iteration number. As shown in Fig. 5(a) the bias-squared value in a full curved, non-TOF scanner has a minimum value after 3 iterations where the reconstruction estimator converges. Limited angle scanners have higher bias-squared values and TOF reconstruction helps to reduce it. Bias-squared values for limited angle scanners also reach a minimum around 3 iterations, but increase rapidly with higher iterations, especially for non-TOF or poor timing resolution. Since MSE is the sum of average biased-squared and average variance, where the variance shows the effect of statistical fluctuations due to the number of counts in data, its value increases as the number of iterations goes up Fig. 5(b) shows that the MSE value increases as the number of iterations goes up and is systematically lower for better TOF timing resolution with larger angular coverage. Based upon this study we next investigate lesion CRC values at iteration number 3 where the bias-squared values reach a minimum.

C. Impact of TOF on CRC in Limited Angle Tomography

Fig. 6 shows CRC values for lesion groups 1, 2, and 3 with 8:1 uptake at $r = 2$ and 6 cm for all curved scanners with 150, 300, 450 and 600 ps TOF as well as non-TOF reconstructions. Results are only shown for those lesion phantom images that were deemed to be relatively artifact-free, i.e. artifacts are minimal and do not affect ability to analyze system performance using our CRC metric. The CRC for non-TOF full curved scanner is the same for all lesion groups, but overall there is some drop in CRC values for 2/3 and 1/2 scanners relative to a full curved scanner. The CRC for group 2 lesions for partial curved scanners is worse than in the full curved scanner due to limited angular coverage effects. For lesions at $r = 6 \text{ cm}$ (Fig. 6(a)) overall CRC is lower relative to lesions at $r = 2 \text{ cm}$ (Fig. 6(b)) for all curved geometries due to partial volume effects arising from parallax error. Similar conclusions are drawn for lesion groups with 4:1 uptake (results not shown here).

D. Impact of 2/3 Curved versus 2/3 Flat Scanner Design

In Fig. 7, we show CRC values for lesion groups 1, 2, and 3 at $r = 2$ and 6 cm for 2/3 curved and flat scanners with 150, 300, 450 and 600 ps TOF reconstruction. The 2/3 flat scanner has lower CRC values for lesions at $r = 2$ cm (near the center) compared to the 2/3 curved scanner due to parallax error effects for events near the center of the scanner arising from the flat detector geometry. For lesions at $r = 6$ cm the 2/3 flat scanner has less drop in CRC for radially outgoing lesions since the parallax error is similar to that seen for lesions near the center. The 2/3 curved scanner, however, does show a drop in the CRC for lesion at $r = 6$ cm relative to lesions near the center due to an increase in parallax error as lesion location increases radially. Similar conclusions are drawn for lesion groups with 4:1 uptake (results not shown here).

E. Impact of Scan Time on Measured CRC Precision in Limited Angle Tomography

We reconstructed the simulated data for shorter scan times and estimated the precision of the measured CRC values as twice the standard deviation of the CRC over the 10 statistically independent data sets. Fig. 8 shows the precision (2σ) of the CRC values for the lesion phantom data set as a function of scan time, averaged over the 9 lesions in lesion groups 1, 2 and 3 (8:1 uptake) for a full curved, non-TOF scanner and 300 ps TOF partial curved or flat scanners. The 2/3 flat scanner with 300 ps TOF timing resolution provides better precision of the CRC values than both the 2/3 and 1/2 curved scanners with 300 ps TOF timing resolution and approaches the precision of the CRC values achieved in the full curved non-TOF scanner. The improved CRC precision in 2/3 flat scanner with 300 ps TOF timing resolution is a result of the more uniform CRC uptake measured in this scanner independent of lesion location. Also, lesions with 4:1 uptake ratio are no longer visible in the images from partial curved or flat scanner geometries with 300 ps TOF timing resolution for scan times of ≤ 5 mins.

F. Impact of a DOI Measuring Detector on CRC in 2/3 Angular Coverage Scanner Geometries

The DOI measuring capability helps maintain the high spatial resolution throughout the FOV of a small diameter scanner by reducing parallax error effects. In Fig. 9, we show the visual improvement in lesion contrast and shape as seen in image for both the 2/3 curved and the 2/3 flat scanner geometries with 300 ps TOF reconstruction. Fig. 10 shows the quantitative effect of implementing a 2-level DOI measuring detector in both the 2/3 curved and the 2/3 flat scanners on lesion CRC values. The average CRC values over 10 independent datasets and over lesion groups 1, 2 and 3 were improved by $\sim 40\%$ for lesions near the center ($r = 2$ cm) and $\sim 20\%$ for lesions near edge ($r = 6$ cm) for the 2/3 flat scanner while they were improved by $\sim 2\%$ for lesions near the center ($r = 2$ cm) and $\sim 21\%$ for lesions near edge ($r = 6$ cm) for the 2/3 curved scanner. Similar improvements were measured for lesions with 4:1 uptake.

G. Impact of Resolution Modeling and a DOI Measuring Detector on image quality metrics in a Full Curved Geometry

Fig. 11 shows reconstructed images for a full curved scanner with no RM and no DOI capability, with RM and no DOI, and with no RM but 2 level DOI. Images indicate that RM or DOI can help improve lesion CRC values in the image. Figs. 12 and 13 show the measured CRC values for these three scanner designs with 8:1 and 4:1 lesion uptake ratios, respectively. Overall CRC values over lesions with 8:1 uptake ratio are slightly higher than the lesions with 4:1 uptake ratio while the trend is similar in both. All CRC values for lesions near the center are the same for all three scanner choices. Both RM and a 2 level DOI capable detector also lead to improved CRC values for lesions radially further away

from the center compared to a standard detector, with a larger increase from RM. However, RM also leads to a different noise structure.

Fig. 14 shows bias-squared and MSE calculated for the full curved scanner with no RM and no DOI capability, with RM and no DOI, and with no RM but 2 level DOI. The 2 level DOI measuring scanner design has the least bias in the reconstructed image. However, the MSE values for all three scanners are similar and increase as a function of iteration number due to increased variance in the image.

IV. Discussion and conclusions

In this work, we have shown the benefit of using TOF information to help mitigate distortions, artifacts, and nonuniformity in the reconstructed tomographic images for a dedicated, breast PET scanner in limited angle geometry. We first characterized the reconstructed system spatial resolution of our modeled system using $1.5 \times 1.5 \times 15 \text{ mm}^3$ LSO crystals. Even though smaller and longer crystals will give improved spatial resolution and system sensitivity we chose this crystal size due to the practicality and cost of manufacturing as well as the potential to achieve better energy and timing resolution with this crystal size.

Without TOF information, the reconstructed image for a warm cylinder with lesions in a limited angle imaging situation has significant non-uniformities. This effect becomes more severe when using a PET scanner with a small ring diameter in order to achieve high sensitivity for quantitative imaging. By employing TOF information, image distortions and non-uniformities can be reduced without any detector rotation. Both MSE and bias-squared metrics as described in equation (1) well characterize overall image uniformity in iterative OSEM tomographic reconstruction as a function of timing resolution and angular coverage. However, note that while the bias-squared and MSE metrics as calculated here are an average over the object in the central transverse image slice, there is a localized effect in the image, which is not fully captured by these metrics. In addition, these metrics were calculated on the interior of the phantom while limited angular coverage leads to some counts being placed outside the object boundary, especially for non-TOF imaging. Better timing resolution with larger angular coverage reduces these artifacts and leads to good CRC values for small lesions. Our results show that the 2/3 curved as well as the 2/3 flat scanners provide good coverage for a full transverse FOV, but the transverse FOV is partially cut-off for the 1/2 curved scanner due to reduced detector angular coverage. For the 2/3 angular coverage scanners, flat detector design relative to the curved design has overall lower CRC values around the central region due to larger parallax error, but these values do not degrade at larger radii as in the curved scanner, thus providing a more uniform CRC estimate over the entire FOV. In addition the 2/3 flat design will have the advantage of incorporation in existing X-ray mammography or tomosynthesis units. Using 2/3 angular coverage, flat scanners with 300 ps TOF timing resolution, a realistic scan time of 5 mins provides similar precision for the CRC values to that achieved with a full curved, non-TOF scanner. A 5 min scan time per breast corresponds to a total scan time of 10 mins for two breasts in the dedicated breast scanner, which is similar to a whole-body scan time of 10–15 mins in a commercial whole-body scanner. However, compared to the clinical scanners the dedicated breast scanner achieves a high spatial resolution ($\sim 1.5 \text{ mm}$ as opposed to $\sim 5 \text{ mm}$), and increased sensitivity due to reduced attenuation of the emitted 511 keV photons since the detectors are placed close to the breast and do not have to travel through the torso as in a clinical scanner. Hence, overall our breast PET scanner design is capable of achieving improved lesion detection and quantification relative to the clinical scanners. In related work [11] we have shown through measurements our ability to design a PET detector using $1.5 \times 1.5 \times 12 \text{ mm}^3$ lutetium-yttrium oxy-orthosilicate (LYSO) crystals and achieving coincidence

timing resolution of 300–400 ps. Note that in performance characteristics such as spatial, energy and timing resolutions LYSO is very similar to LSO, though the sensitivity is slightly reduced ($< 10\%$). Since our simulation results show good imaging performance for 2/3 angular coverage scanners with TOF timing resolution less than 600 ps, we therefore think that a 2/3 angular coverage scanner can be designed to provide accurate tomographic images for breast imaging.

To further improve the performance of the limited angle scanner design by reducing parallax error, a DOI capable detector and/or RM were also evaluated here. A 2 level DOI measuring detector helps overcome parallax error effects and increases CRC values, but shows the most improvement in the 2/3 flat scanner design. RM on the other hand has a larger impact on the CRC in the curved scanner geometry and produces similar CRC as the 2 level DOI detector in a full curved geometry. The bias-squared value is higher with RM but reduced variance in the image leads to similar MSE with RM or 2 level DOI in a full curved scanner. Future work will involve incorporation and testing of RM techniques in the limited angle geometry scanner design and using the MSE metric to better evaluate the impact of DOI and RM in the partial ring scanner designs.

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Fig. 1.

The scanner setup for full curved, 2/3 (120° in-plane angular coverage) curved, and 1/2 (90° in-plane angular coverage) curved as well as 2/3 flat (120° in-plane angular coverage). Both the ring diameter (detector separation for 2/3 flat) and the axial length are 15 cm for the scanners.

Fig. 2.

Simulated lesion phantom with 25, 5 mm diameter spheres at multiple radial positions ($r = 0, 2, 4$ and 6 cm) with 4:1 and 8:1 activity uptake ratios with respect to the background. The cylinder diameter and axial length were 14 cm. Note that the partial curved detectors as shown here provide a schematic view of their location relative to the lesion phantom and all subsequent studies have the same detector placement.

Fig. 3. Reconstructed system spatial resolution (FWHM) for point source in air calculated using NEMA NU 4-2008 procedure and for point source in warm background using NEMA NU 4-2008 followed by a Gaussian fit with an offset due to a warm background. Results are shown for the full curved scanner geometry.

Fig. 4.

Reconstructed lesion phantom images of a central transverse slice for all scanner geometries with (a) Non-TOF, (b) 600 ps TOF, (c) 450 ps TOF, (d) 300 ps TOF and (e) 150 ps TOF reconstructions. Within each row the four images from left to right are for full curved, 2/3 curved, 1/2 curved and 2/3 flat scanner geometries.

Fig. 5. The (a) bias-squared and (b) MSE of reconstructed lesion phantom images plotted as a function of iteration number for Non-TOF, 600, 300 and 150ps TOF with various angular coverage of curved and flat scanners.

Fig. 6. CRC values for lesion groups 1, 2 and 3 (8:1 uptake ratio) at (a) $r = 2$ cm and (b) $r = 6$ cm for all curved geometries for 150, 300, 450, 600 ps TOF as well as non-TOF reconstruction. Results are shown only for those images that were deemed to be relatively artifact free.

Fig. 7. CRC values for lesion groups 1, 2 and 3 (8:1 uptake ratio) at (a) $r = 2$ cm and (b) $r = 6$ cm for 2/3 curved versus 2/3 flat scanner geometries for 150, 300, 450, 600 ps TOF reconstruction.

Fig. 8.

The precision of measured CRC values estimated over 10 independent lesion phantom data sets as a function of scan time, which is averaged over 9 lesions in lesion group 1, 2 and 3. Results are shown for a full curved scanner with non-TOF reconstruction and 2/3 curved, 1/2 curved, and 2/3 flat scanners with 300 ps TOF reconstruction. Note that 2/3 flat scanner has off-set time stamps relative to curved scanners and x-axis is in log-scale.

Fig. 9. Central transverse slice of the reconstructed image for (a) 2/3 curved and (b) 2/3 flat scanners with 300 ps TOF using (left) a non-DOI measuring detector and (right) a 2-level DOI measuring detector.

Fig. 10.

CRC values for lesion groups 1, 2 and 3 (8:1 uptake ratio) at (a) $r = 2$ cm and (b) $r = 6$ cm (bottom) for the 2/3 curved and the 2/3 flat scanners with 300 ps TOF and using a non-DOI or 2-level DOI measuring detector.

Fig. 11. Central transverse slice of the reconstructed images for a full curved geometry with 300 ps TOF scanner. Images are shown for (left) a standard scanner (no RM and no DOI), (middle) a scanner with RM but no DOI, and (right) scanner with 2-level DOI, but no RM.

Fig. 12. CRC values after incorporating RM and DOI detector for lesion groups 1, 2 and 3 (8:1 uptake ratio) at (a) $r = 2$ cm and (b) $r = 6$ cm for a full curved geometry and 300 ps TOF scanner.

Fig. 13. CRC values after incorporating RM and DOI detector for lesion groups 5, 6 and 7 (4:1 uptake ratio) at (a) $r = 2$ cm and (b) $r = 6$ cm for a full curved geometry and 300 ps TOF scanner.

Fig. 14.

The (a) bias-squared and (b) MSE of reconstructed lesion phantom images plotted as a function of iteration number after incorporating RM and DOI detector for a full curved geometry and 300 ps TOF scanner.