Published in final edited form as:

IEEE Trans Nucl Sci. 2010 October 1; 57(5): 2497–2503. doi:10.1109/TNS.2010.2049121.

Imaging studies for evaluating impact of position sampling techniques in PET scanners

Suleman Surti, IEEE[Senior Member],

Department of Radiology, University of Pennsylvania, Philadelphia, PA 19104, USA (telephone: 215-662-7214, surti@mail.med.upenn.edu).

Matthew E. Werner, and

Department of Radiology, University of Pennsylvania, Philadelphia, PA 19104, USA (telephone: 215-662-7217, Matt.Werner@uphs.upenn.edu).

Joel S. Karp, IEEE[Senior Member]

Departments of Radiology and Physics, University of Pennsylvania, Philadelphia, PA 19104 USA (telephone: 215-662-3073, joelkarp@mail.med.upenn.edu).

Abstract

Previously we have evaluated two crystal calibration techniques that can be applied to pixelated detector designs to improve system spatial resolution without detector motion. The inter-crystal positioning technique utilizes sub-sampling in the crystal flood map to better sample the Compton scatter events in the detector. The Compton scatter rejection technique, on the other hand, rejects those events that are located further from individual crystal centers in the flood map. Here we performed imaging studies with a Mini Deluxe hot rod phantom and a hot sphere phantom (sphere diameters of 4.95 and 7.86-mm with 6:1 uptake relative to background) using the standard crystal calibration technique, as well as the inter-crystal and Compton rejection calibration techniques. Our results show improved separation of 1.6-mm diameter hot rods with the two new crystal calibration techniques that is consistent with improved spatial resolution. For the hot sphere phantom the contrast recovery is improved with both the inter-crystal and Compton rejection calibration techniques over the standard calibration technique. The only drawback of the inter-crystal calibration technique is the increase in the number of possible lines-of-response (LORs) (factor of 16) that may slow image reconstruction. With the Compton rejection calibration technique, loss of counts leads to increased noise in the images.

I. INTRODUCTION

In recent years the trend in PET detector design has been towards the use of small cross-section crystals in some form of a light-sharing detector [1]. Theoretically, assuming very high scintillation photon statistics, the intrinsic resolution due to detector effects is $d/2$ for a crystal of cross-section size, d . However, the discrete (under) sampling due to fixed-size pixelated detectors leads to a convolution of the intrinsic resolution with the detector size, where the detector size limits the scanner spatial resolution due to under sampling [2]. Even in clinical scanners where the photon non-collinearity and initial ^{18}F positron range limit the reconstructed spatial resolution to $>2\text{-mm}$ [3], the use of 4-mm wide crystals provides the limit on the reconstructed spatial resolution due to under sampling. In order to improve the sampling in pixelated detectors and achieve results closer to the theoretical estimates, researchers in the past have developed techniques such as “wobbling” which involves a mechanical movement of the detector ring while performing patient scans [4–6]. Spatial resolution can also be improved (without any hardware change) by characterizing the detector response [7], followed by modeling this response in the system matrix for image reconstruction. Results have shown improvements in the resultant PET images [7].

In earlier work [8] we investigated two crystal calibration techniques that can be applied to pixelated detector designs to improve system spatial resolution without detector motion. In Fig. 1 we show schematically the three crystal calibration techniques that we investigated. In standard crystal calibration (Fig. 1A) the crystal boundary in the look-up overlaps with the physical crystal boundary leading to a region-of-interest (ROI) size that is the same as the crystal size. With the inter-crystal calibration technique (Fig. 1B) the ROI size is half the crystal size and neighboring ROIs are adjacent to each other with no gaps. Finally, with the Compton scatter rejection technique, the ROI sizes are reduced such that neighboring ROIs are not adjacent to each other (Fig. 1C). The ROI size is variable and will affect the number of counts collected for the study.

In Fig. 2 we show a practical implementation of these three calibration techniques on measured data from the small animal *APET* scanner [9, 10]. Fig. 2A shows a 2D position flood maps over a large portion of the scanner, while Figs. 2B–E show a single PMT region for clarity. Note that *APET* has a continuous, cylindrical detector with $14,456\ 2\times 2\times 10\text{-mm}^3$ LYSO crystals coupled through an annular lightguide to a hexagonal array of 19-mm diameter PMTs in a pixelated Anger-logic detector design [10]. Thus, while there is some pincushion effect near a PMT center, the only significant distortion is near the detector boundaries at the axial edges (see Fig. 2A), which is also where it becomes more difficult to reliably implement the new crystal calibration techniques. For standard calibration, the individual crystal peaks are identified in the flood map followed by a *Vornoi* decomposition to mark the boundary region or ROI around each crystal (Fig. 2C). The inter-crystal technique makes use of good crystal separation in the detector flood map to increase the number of bins generated in the look-up table that maps measured interaction positions to individual crystals (Fig. 2D). In practice, virtual crystal peak locations are introduced exactly halfway between the real crystal peaks, followed by *Vornoi* decomposition as performed earlier for standard calibration. In the inter-crystal technique, the ROI bin sizes are reduced to about half their size and an additional bin is introduced between adjacent crystals, effectively quadrupling the number of crystals and increasing the number of lines-of-response (LORs) by a factor of 16. The Compton rejection technique (Fig. 2E), on the other hand, reduces the size of the individual ROIs centered over each crystal in the flood map and rejects any events that fall outside the ROI. In our implementation, the ROI boundary is determined by selecting a fraction of the counts that are to be retained and are clustered close the ROI center. With the Compton rejection technique there will therefore be a trade-off between spatial resolution and sensitivity.

Our previous work demonstrated an improvement in the measured spatial resolution for a point source in air using these new crystal calibration techniques [8]. In particular, when applied to the small animal *APET* scanner [9, 10], the standard NEMA [11] spatial resolution value improves from 1.9/4.7-mm (fwhm/fwtm) to 1.6/4.4-mm with the inter-crystal technique and 1.7/4.2-mm with the Compton rejection technique with the rejection of 50% of the events. In comparison [12], a mechanical wobbling technique implemented on the small animal MicroPET scanner [13, 14] which also uses 2-mm wide crystals, showed a similar improvement in the reconstructed spatial resolution, improving from 1.8-mm (fwhm) without wobbling to 1.6-mm (fwhm) with wobbling.

In Table I we summarize the advantages and disadvantages of these crystal calibration techniques in addition to detector wobbling. As shown in the table, the inter-crystal technique and detector “wobbling” techniques will result in increased sinogram matrix size due to increased LOR sampling. Normalization data for the scanner in these situations will have to be acquired with a much higher number of counts than the standard acquisitions due to the increased number of sinogram bins. Our goal for this work is to perform imaging experiments on the *APET* scanner using a hot rod phantom and a small lesion phantom to

demonstrate improvements in image quality, and possible drawbacks, that could be achieved with the inter-crystal and Compton rejection calibration techniques over the standard crystal calibration technique. In this work we will limit the Compton rejection calibration technique to the one where the ROI size is chosen to reject 50% of the events. While the Compton rejection technique can be applied with a higher or lower percentage of event rejection, for ease of testing and measurements we restricted ourselves to a 50% rejection rate. Based on LYSO stopping power, we calculate that about 65% of interacting 511 keV photons deposit their energy within the 2-mm wide crystal they first enter. After accounting for some event mis-positioning in the detector due to light sharing effects, it therefore seems reasonable to use a 50% rejection rate in order to reduce Compton scattered events in the detector. All measurements were performed with the scanner default energy window that set at 385–665-keV.

II. METHODS

A. Data acquisition

Phantom data were acquired using a Mini Deluxe hot rod phantom (Data Spectrum Corporation, Hillsborough, NC) as well as a hot sphere phantom. The hot rod phantom (Fig. 3) consists of a 4.4-cm diameter by 3.4-cm long cylinder with a hot rod insert. The hot rod insert consists of six segments with rods of diameter 1.2, 1.6, 2.4, 3.2, 4.0, and 4.8-mm within each segment, and a rod spacing that is twice the rod diameter. The hot sphere phantom consists of a 7.6-cm diameter by 5.0-cm long cylinder that has four spheres of internal diameters 4.95, 4.95, 7.86, and 7.86-mm. The spheres were placed so that their centers were aligned in the same plane at 4.0-cm from one end of the cylinder. For both measurements data were collected with a total of 600 μCi of ^{18}F -FDG present in the phantom; for the hot rod setup there was no activity in the background, while for the hot sphere setup the sphere activity concentration was 6:1 with respect to the background. Scan time was fixed at 15 minutes, and data were collected in list-mode format with standard, inter-crystal, and Compton rejection crystal calibration techniques.

B. Image reconstruction and analysis

For image reconstruction, attenuation was estimated analytically for the phantoms by drawing a cylindrical ROI around the phantom edges and assuming uniform water attenuation within it. Scatter correction was performed using the model-based single scatter simulation [15–17]. Randoms were estimated using a delayed window technique. Image reconstruction was performed using two different reconstruction algorithms. Pre-corrected data were first reconstructed using the analytical 3D-FRP [18] algorithm. In addition, an un-relaxed OSEM list-mode reconstruction algorithm (33 subsets) with built-in system corrections for attenuation, scatter, and randoms (blob basis functions but no detector response modeling) was also used [19, 20]. The analytical 3D-FRP algorithm was chosen because it is more linear, and so perhaps provides a better representation of the spatial resolution improvement due to the crystal calibration techniques. On the other hand an iterative reconstruction algorithm is routinely used on most modern PET scanners (including *APET*) due to its ability to achieve high image signal-to-noise ratio, and so we wanted to test the influence of crystal calibration techniques with such an algorithm as well.

For the hot rod phantom line profiles were drawn to demonstrate discrimination of the hot rods. For the hot sphere phantom, mean signal count density (C_H) and its standard deviation (σ_H) were determined for circular regions of interest (ROIs) drawn with varying size diameters in the slice containing the spheres and centered over the hot spheres. A 1-cm diameter background ROI was also drawn in this slice centered in the middle of the slice to obtain the mean background count density (C_B) and its standard deviation (σ_B). The sphere

contrast recovery coefficient (*CRC*) and image pixel noise (*Noise*) were then determined from:

$$CRC = 100\% \times \frac{\left(\frac{C_H}{C_B} - 1\right)}{\left(\frac{a_H}{a_B} - 1\right)}, \quad Noise = \frac{\sigma_B}{C_B}$$

where a_H and a_B are the activity concentrations in the spheres and background (6 and 1), respectively.

III. RESULTS

A. Hot rod phantom

In Fig. 4 we show reconstructed images for the hot rod phantom as a function of a few iteration numbers for the list-mode iterative reconstruction (using standard calibration). We picked iteration number 5 as being representative of the iterative reconstruction (later in Fig. 8 we show that maximum image contrast is achieved at or near iteration 5), and all subsequent results use this iteration number.

In Fig. 5 we show a central transverse slice of the reconstructed images using the three different crystal calibration techniques obtained with the iterative list-mode image reconstruction and the analytical 3D-FRP reconstruction. These data were acquired with high count statistics where the noise in the 3D-FRP algorithm is not a limiting factor. To better demonstrate the differences in the images due to the crystal calibration techniques, in Fig. 6A we show a profile drawn through a row of three 2.4-mm diameter and five 1.6-mm diameter hot rods. Fig. 6B shows another profile drawn through another row of 1.6-mm diameter hot rods. While these profiles may indicate some improved performance with the new calibration techniques, additional measurements with the hot sphere phantom are needed.

B. Hot sphere phantom

In Fig. 7 we show a transverse slice through the centers of the spheres in the reconstructed images of the sphere phantom. Images are shown for data acquired with the three calibration techniques and reconstructed with the iterative reconstruction algorithm. While, the two sets of spheres are clearly visible in all three images, the reduced number of counts (50%) in the Compton rejection calibration leads to a noisier image.

Quantitatively, in Fig. 8 we plot the CRC versus noise relationship for images from these three calibration techniques for ten iterations of the list-mode reconstruction algorithm. The estimated error in the CRC values is less than 1%. For all three crystal calibration techniques, a maximum CRC value is achieved at or near iteration number 5. For both sphere sizes, Compton rejection and inter-crystal calibrations lead to higher CRC values relative to the standard calibration. Between the Compton rejection and inter-crystal calibration techniques, the results vary depending on sphere size. For the smaller 4.95-mm diameter sphere, Compton rejection leads to a higher CRC value while for the 7.86-mm diameter sphere, inter-crystal leads to a higher CRC value. This may be explained by the narrower fwhm values of the Compton rejection point-spread-function (4.2-mm as opposed to 4.4-mm) [8], where the short tails provide an improved estimation of mean counts within a smaller region, and so lead to higher CRC for small spheres. Also, as noted earlier, the reduced number of counts in the Compton rejection calibration data leads to higher pixel

noise at each iteration number (curve shifted toward higher noise relative to the standard and inter-crystal curves).

In Table II we summarize the CRC and Noise values achieved for the two sphere sizes with the three different calibration techniques and performing analytical 3D-FRP image reconstruction. In addition, we also show results from Fig. 8 for iterative reconstruction after 5 iterations (close to maximum CRC is achieved for the spheres). Comparing the 3D-FRP results to iterative reconstruction results for each crystal calibration, we obtain higher CRC values with iterative reconstruction. Using fewer iterations than five (eg. three) does not reduce CRC significantly but improved Noise values could be achieved. Also, with the 3D-FRP reconstruction the overall trend in CRC values between the three calibration techniques is similar to what we observed with the iterative list-mode reconstruction algorithm.

IV. DISCUSSION AND CONCLUSION

Through phantom measurements acquired on the small animal *APET* scanner we evaluated the impact of the inter-crystal and Compton rejection calibration techniques on the reconstructed images. Our results indicate that hot rod discrimination is improved with these new crystal calibration techniques over the standard crystal calibration techniques for rods with 1.6-mm diameter. This result is consistent with the improved spatial resolution of 1.6–1.7-mm (fwhm) we measured previously with these calibration techniques compared to 1.9-mm after standard calibration [8]. The CRC values for the two spheres in the lesion phantom also show improvements with the new crystal calibration techniques. These results were verified by first using an analytical 3D-FRP reconstruction algorithm followed by an iterative reconstruction algorithm which generally performed slightly better.

As expected, the noise in the images acquired with the Compton rejection crystal calibration technique was higher compared to the standard and inter-crystal calibration techniques due to a loss of 50% of the collected counts. However, by using smooth, Kaiser-Bessel (blob) basis function [21] in an iterative image reconstruction algorithm we can better control the effect of increased noise in these images.

For the inter-crystal calibration technique, the primary concern is the impact on image reconstruction speed due to a factor of 16 increase in the number of possible LORs. While we did not evaluate its use in this study, an iterative LOR sinogram based reconstruction algorithm represents the most commonly used reconstruction technique in clinical and small animal imaging PET scanners. A factor of 16 increase in the number of LORs can therefore significantly increase the reconstruction times. An alternative to an iterative LOR sinogram based reconstruction algorithm is a list-mode iterative reconstruction algorithm whose reconstruction time increases as the number of collected events increases and is independent (to a large degree) of the number of LORs. This was one of the reasons that we evaluated the use of a list-mode iterative reconstruction algorithm in this work. For the iterative list-mode reconstruction algorithm implemented in this work, we benchmark a reconstruction time of 45 seconds/iteration/1 million events with the inter-crystal technique as opposed to 35 seconds/iteration/1 million events with standard calibration (parallel processes running on a dual 2.0 GHz quad-core AMD Opteron processor). This indicates that the inter-crystal calibration technique can be feasible to use when implemented together with an iterative list-mode reconstruction algorithm with a 30% increase in reconstruction.

Our imaging experiments show the improved imaging capability in terms of hot rod discrimination and small sphere uptake estimation achieved by the inter-crystal and Compton rejection crystal calibration techniques. This improvement was achieved with either some increased software overhead in image reconstruction (inter-crystal) times or loss

of counts (Compton rejection). Traditionally, more significant imaging improvements are achieved by using smaller crystals (increased cost and complexity), and similar imaging improvements are achieved after employing detector motion (increased mechanical complexity). While the new crystal calibration techniques were implemented on the *APET* scanner that uses a pixelated Anger-logic detector, these techniques can in principle also be implemented on other pixelated detector designs. However, the edges in a block detector design will be more problematic to deal with due to poorer crystal discrimination near the detector edge. The relative impact of these techniques on reconstructed images from other detector designs may therefore vary, depending on the crystal type, degree of crystal separation achieved in the flood images, and the operating energy window for accepted events in the scanner. Also, while beyond the scope of this investigation, it may be of interest to compare the imaging performance of these new calibration techniques to the standard calibration, with and without modeling the detector response in the reconstruction algorithm [7].

We envision using the inter-crystal calibration technique in general small animal imaging situations where both improved uptake estimation and reduced noise in the image are important factors to consider. The Compton rejection technique could be used in brain imaging studies in small animals where a high specific uptake radiotracer is used to image small regions such as brain receptors.

Acknowledgments

We would like to thank Eric Blankemeyer (University of Pennsylvania) for help with data collection and Ryan Scheuermann (University of Pennsylvania) for the early work performed with the new crystal calibration techniques.

This work was supported by the National Institutes of Health under Grant Nos. R21-EB008142, R01-CA113941, and R01-EB009056.

REFERENCES

1. Muehllehner G, Karp JS. Positron emission tomography. *Phys Med Biol.* 2006; vol. 51:R117–R137. [PubMed: 16790899]
2. Huang SC, Hoffman EJ, Phelps ME, Kuhl DE. Quantitation in Positron Emission Computed-Tomography .3. Effect of Sampling. *J Comput Assist Tomogr.* 1980; vol. 4:819–826. [PubMed: 6971301]
3. Cherry, SR.; Sorenson, JA.; Phelps, ME. *Physics in Nuclear Medicine.* Philadelphia: Saunders; 2003.
4. Bohm C, Eriksson L, Bergstrom M, Litton J, Sundman R, Singh M. A computer assisted ring detector positron camera system for reconstruction tomography of the brain. *IEEE Trans Nucl Sci.* 1978; vol. NS-25:624–637.
5. Brooks RA, Sank VJ, Talbert AJ, Chiro GD. Sampling requirements and detector motion for positron emission tomography. *IEEE Trans Nucl Sci.* 1979; vol. NS-26:2760–2763.
6. Tanaka E, Nohara N, Yamamoto M, Tomitani T, Murayama H, Ishimatsu K, Takami K. 'Positology' - The search for suitable detector arrangements for a positron ECT with continuous rotation. *IEEE Trans Nucl Sci.* 1979; vol. NS-26:2728–2731.
7. Panin VY, Kehren F, Michel C, Casey M. Fully 3-D PET reconstruction with system matrix derived from point source measurements. *IEEE Trans Med Imag.* 2006; vol. 25:907–921.
8. Surti S, Scheuermann R, Werner ME, Karp JS. Improved spatial resolution in PET scanners using sampling techniques. *IEEE Trans Nucl Sci.* 2009; vol. 56:596–601. [PubMed: 19779586]
9. Surti S, Karp JS, Perkins AE, Freifelder R, Muehllehner G. Design evaluation of A-PET: A high sensitivity animal PET camera. *IEEE Trans. Nucl. Sci.* 2003; vol. 50:1357–1363.

10. Surti S, Karp JS, Perkins AE, Cardi CA, Daube-Witherspoon ME, Kuhn A, Muehllehner G. Imaging performance of A-PET: A small animal PET camera. *IEEE Trans. Med. Imag.* 2005; vol. 24:844–852.
11. NEMA Association. Rosslyn, VA: National Electrical Manufacturers Association; Performance Measurements of Positron Emission Tomographs. NEMA Standards Publication NU 2-2001 2001
12. Chatzioannou A, Silverman RW, Meadors K, Farquhar TH, Cherry SR. Techniques to improve the spatial sampling of microPET - A high resolution animal PET tomograph. *IEEE Trans Nucl Sci.* 2000; vol. 47:422–427.
13. Chatzioannou AF, Cherry SR, Shao YP, Silverman RW, Meadors K, Farquhar TH, Pedarsani M, Phelps ME. Performance evaluation of microPET: A high-resolution lutetium oxyorthosilicate PET scanner for animal imaging. *J Nucl Med.* 1999; vol. 40:1164–1175. [PubMed: 10405138]
14. Cherry SR, Shao Y, Silverman RW, Meadors K, Siegel S, Chatzioannou A, Young JW, Jones W, Moyers JC, Newport D, Boutefnouchet A, Farquhar TH, Andreaco M, Paulus MJ, Binkley DM, Nutt R, Phelps ME. MicroPET: a high resolution PET scanner for imaging small animals. *IEEE Trans Nucl Sci.* 1997; vol. 44:1161–1166.
15. Watson CC. New, faster, image-based scatter correction for 3D PET. *IEEE Trans Nucl Sci.* 2000; vol. 47:1587–1594.
16. Ollinger JM. Model-based scatter correction for fully 3D PET. *Phys Med Biol.* 1996; vol. 41:153–176. [PubMed: 8685253]
17. Accorsi R, Adam LE, Werner ME, Karp JS. Optimization of a fully 3D single scatter simulation algorithm for 3D PET. *Phys Med Biol.* 2004; vol. 49:2577–2598. [PubMed: 15272675]
18. Matej S, Lewitt RM. 3D-FRP: Direct fourier reconstruction with Fourier reprojection for fully 3-D PET. *IEEE Trans. Nucl. Sci.* 2001; vol. 48:1378–1385.
19. Popescu, LM. Iterative image reconstruction using geometrically ordered subsets with list-mode data. presented at 2004 IEEE Nuclear Science Symposium and Medical Imaging Conference; Rome, Italy. 2004.
20. Popescu, LM.; Lewitt, RM. Tracing through a grid of blobs. presented at 2004 IEEE Nuclear Science Symposium and Medical Imaging Conference; Rome, Italy. 2004.
21. Lewitt RM. Alternatives to Voxels for Image Representation in Iterative Reconstruction Algorithms. *Phys Med Biol.* 1992; vol. 37:705–716. [PubMed: 1565698]

Fig. 1.

Schematic picture showing the three crystal calibration techniques as applied to a 5×5 array of crystals. The thick lines show the ROI drawn for look-up table generation. (A) Standard calibration where the ROI is centered over each crystal and is equal to the crystal size. (B) Inter-crystal calibration technique where an ROI is centered over each crystal and is equal to half the crystal size. An additional ROI is placed between two such centered ROIs in each direction, leading to a quadrupling in the number of crystal ROIs generated. (C) Compton scatter rejection technique where the ROIs are centered over each crystal and are smaller in size. Events that do not lie within an ROI are now rejected.

(A)

(B)

(C)

(D)

(E)

Fig. 2.

Schematic picture of drawing regions around individual crystals during crystal calibrations. (A) Flood map over a portion of the *APET* detector. Each hexagonal region shows the counts distributed over the crystals below a single PMT. The hexagonal pattern is due to the non-linear positioning algorithm that uses signals from a seven PMT cluster for positioning an event (the PMT which collects the maximum signal and its six immediate neighboring PMTs). The vertical direction is the full axial extent of the *APET* scanner, which shows the impact of detector edge on crystal discrimination. (B) A flood map over a portion of a single PMT region showing events within individual crystal. (C) Standard calibration technique — All measured events are placed at the physical center (crystal position on the scanner) of an

ROI. The ROI regions are centered over individual crystals and with spacing equal to the crystal pitch. (D) Inter-crystal positioning technique — All measured events are placed at the physical center of an ROI (every half crystal position on the scanner). The ROI regions are centered over individual crystals with spacing that is half the crystal pitch leading to four times the number of ROIs as in the standard calibration technique. (E) Compton scatter rejection technique — All measured events are placed at the physical center (crystal position on the scanner) of a ROI which are now equal or smaller than the crystal pitch. The events outside these regions are rejected.

Fig. 3.
A Mini Deluxe hot rod phantom that was used in the measurements.

Fig. 4.

Transverse slices from reconstructed images of the hot rod phantom as a function of iteration number after list-mode iterative reconstruction.

Fig. 5.

Transverse slices from reconstructed images of the hot rod phantom for the three different crystal calibration techniques (standard, Compton rejection, and inter-crystal) and two image reconstruction algorithms (iterative list-mode and analytical 3D-FRP). The iterative reconstruction images are after five iterations.

Fig. 6. Profiles drawn through the hot-rod phantom images for the three crystal calibration techniques. Results are shown for iterative list-mode image reconstruction (5 iterations), and the profile is drawn through (A) three 2.4-mm diameter and five 1.6-mm diameter rods, and (B) two 4.8-mm diameter and five 1.6-mm diameter rods. Each profile was normalized to the total counts present in that profile.

Fig. 7. Transverse slices from reconstructed images of the hot sphere phantom for the three different crystal calibration techniques (standard, Compton rejection, and inter-crystal). Images were reconstructed using the iterative list-mode algorithm (5 iterations).

Fig. 8. CRC versus Noise plots for the (A) 4.95-mm diameter and (B) 7.86-mm diameter spheres in the hot sphere phantom. The sphere uptake was 6:1 with respect to background. Each point along the curves represents increasing number of iterations, starting with iteration 1 in the lower left corner to iteration 10 in the upper right corner. The estimated error in the CRC values is less than 1%.

TABLE 1

Summary of different crystal calibration techniques as well as detector wobbling. The variable d represents the crystal size.

	Bin size	Bin spacing	Advantages	Disadvantages
Standard bins	$\sim d$	$\sim d$	N/A	N/A
Compton rejection bins	$\sim d/2$	$\sim d$	Improved spatial resolution	Loss of counts
Inter-crystal bins	$\sim d/2$	$\sim d/2$	Improved spatial resolution	Large sinogram matrix
Detector wobbling	$\sim d$	$\sim d/2$	Improved spatial resolution	Large sinogram matrix Detector motion

Summary of CRC and Noise values for the two sphere sizes as a function of crystal calibration technique after using 3D-FRP reconstruction algorithm as well as five iterations of the list-mode reconstruction algorithm.

TABLE II

		Standard calibration		Compton rejection calibration		Inter-crystal calibration	
		CRC	Noise	CRC	Noise	CRC	Noise
3D-FRP reconstruction	4.95-mm diameter sphere	30%	0.14	38%	0.21	35%	0.19
	7.86-mm diameter sphere	52%	0.14	59%	0.21	60%	0.19
Iteration 5 of list-mode reconstruction	4.95-mm diameter sphere	37%	0.16	44%	0.33	42%	0.15
	7.86-mm diameter sphere	55%	0.16	60%	0.33	64%	0.15