Published in final edited form as:

Phys Med Biol. 2016 July 21; 61(14): 5443–5455. doi:10.1088/0031-9155/61/14/5443.

Parallax error in long-axial field-of-view PET scanners—a simulation study

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Abstract

There is a growing interest in the design and construction of a PET scanner with a very long axial extent. One critical design challenge is the impact of the long axial extent on the scanner spatial resolution properties. In this work, we characterize the effect of parallax error in PET system designs having an axial FOV of 198 cm (total-body PET scanner) using fully-3D Monte Carlo simulations. Two different scintillation materials were studied: LSO and LaBr₃. The crystal size in both cases was 4×4×20 mm³. Several different depth-of-interaction (DOI) encoding techniques were investigated to characterize the improvement in spatial resolution when using a DOI capable detector. To measure spatial resolution we simulated point sources in a warm background in the center of the imaging FOV, where the effects of axial parallax are largest, and at several positions radially offset from the center. Using a line-of-response based OSEM reconstruction algorithm we found that the axial resolution in an LSO scanner degrades from 4.8 mm to 5.7 mm (FWHM) at the center of the imaging FOV when extending the axial acceptance angle (α) from $\pm 12^\circ$ (corresponding to an axial FOV of 18 cm) to the maximum of $\pm 67^\circ$ —a similar result was obtained with LaBr₃, in which the axial resolution degraded from 5.3 mm to 6.1 mm. For comparison we also measured the degradation due to radial parallax error in the transverse imaging FOV; the transverse resolution, averaging radial and tangential directions, of an LSO scanner was degraded from 4.9 mm to 7.7 mm, for a measurement at the center of the scanner compared to a measurement with a radial offset of 23 cm. Simulations of a DOI detector design improved the spatial resolution in all dimensions. The axial resolution in the LSO-based scanner, with $\alpha = \pm 67^\circ$, was improved from 5.7 mm to 5.0 mm by incorporating a two-layer DOI detector. These results characterize the maximum axial blurring for a fully open 2 m long PET scanner and demonstrate that large sensitivity gains are possible with a modest reduction in resolution when using current clinical detector technology with no DOI capability.

1. Introduction

There is a renewed interest in developing clinical positron emission tomography (PET) scanners that have a large field-of-view (FOV) in the axial direction, potentially to the extent that the entire body is covered (Badawi *et al* 2000, Cherry 2006, Wong *et al* 2007, Eriksson *et al* 2011, Poon *et al* 2012, Crespo *et al* 2012). Such a long scanner design could be capable of very high system sensitivity, due to the large increase in solid angle coverage, and it would also enable applications that are not possible with the sequential whole-body imaging

approaches available using existing commercial PET/CT systems (Price *et al* 2014, Zhang *et al* 2014), which have an axial FOV ranging from 18 cm – 22 cm (Surti *et al* 2007, Bettinardi *et al* 2011, Jakoby *et al* 2011). Simultaneous high-sensitivity imaging of the entire body has the potential to fundamentally change how PET is currently used—from the choice of isotope and its administered activity, to the acquisition of truly 4-D PET data and reconstruction techniques optimized for its analysis, which will expand imaging applications and incorporate new patient populations. There is no consensus for the optimal axial length that will be needed for simultaneous total-body imaging—it will of course depend on the application—however it is clear that this paradigm will require that a significant portion of the body be imaged simultaneously.

There are many technical challenges that affect the feasibility of a scanner design with a long axial extent (in addition to cost). Previously, two prototype systems with an extended axial FOV have been built: a LSO-based panel detector system having a length of 53 cm (Conti *et al* 2006) and a BGO-based 2-D acquisition system with an axial extent of 68.5 cm (Watanabe *et al* 2004). These systems had high sensitivity but their count-rate performance was limited due to compromises made in the system electronics, and in Watanabe *et al* 2004 the use of BGO scintillator. In addition, both of these prototype systems failed to demonstrate unique imaging applications. Currently there are several groups actively investigating the design and construction of a long-axial FOV system with improved performance (Badawi *et al* 2013, Cherry *et al* 2013, Crespo 2012). There have been numerous simulation studies investigating the gain in imaging performance with increasing axial FOV using the noise equivalent count (NEC) metric (Badawi *et al* 2000, Poon *et al* 2012, MacDonald *et al* 2011, Surti *et al* 2013, Isnaini *et al* 2014). The conclusions from some of these studies also aimed to address the cost-effectiveness of extending the axial FOV, but in general all showed large gains in system NEC with increasing axial FOV, with Poon *et al* 2012 reporting an NEC gain of 25–31 $\times$, compared to standard length axial FOV systems, with 2 cm thick detectors and 2 m of axial extent. A major limitation of some of these studies using NEC to optimize PET system design is that improvements in image quality due to time-of-flight (TOF) and depth-of-interaction (DOI) information cannot be measured. In addition, NEC is known to have limitations in predicting the image SNR when using advanced 3D iterative reconstruction algorithms, which are now commonly used clinically (Chang *et al* 2012). TOF-PET has improved many components of the imaging process, leading to a significant improvement in the quality of clinical PET images (Karp *et al* 2008, Surti 2015), and it is expected that TOF gains will also translate to improved image quality in long-axial scanner designs. Degradations to image spatial resolution from DOI effects may have a heightened impact in long-axial scanners. To fully exploit the sensitivity gains of extended axial length, and greater solid angle coverage, the axial acceptance angle will be large and cause a parallax error in the axial direction (a fully-3D 2 m long scanner with a 85 cm diameter will have an axial acceptance angle $\pm 67^\circ$). This parallax error will be most pronounced in the center of the axial and transverse imaging FOV, and in addition to an inherent radial parallax error associated with the ring geometry, it has been suggested that PET scanner designs with long-axial FOVs may have worse spatial resolution compared to more traditional scanner designs having a shorter axial FOV (Berg *et al* 2016, Zhang *et al* 2016).

In a recent paper by Surti and Karp 2015, a simulation study was presented that used image-based metrics to investigate the impact of detector performance on a long-axial FOV scanner having an axial FOV of 72 cm; this study included analysis of TOF resolution and a 2-layer DOI detector on image quality. It was hypothesized that detector configurations capable of measuring both TOF and DOI, as demonstrated in studies such as (Schaart *et al* 2009, Wiener *et al* 2013, Schmall *et al* 2014, Yeom *et al* 2014,), would improve imaging performance particularly for the simulated 72 cm long scanner design. Results from the study showed large gains in task-based image quality metrics (lesion detectability and contrast measurement) with better TOF performance but no significant improvement when using a 2-layer DOI detector. The results of this prior study were based on imaging 1 cm diameter lesions with low contrast (3–1 lesion to background) in a scanner design having an axial FOV of 72 cm, and therefore may not have been sensitive to improved spatial resolution resulting from the DOI measurement. However, it is also possible that degradations to spatial resolution from parallax error in the axial direction are relatively small, and do not significantly impact image quality when using modern, fully-3D image reconstruction algorithms. Therefore in this work we specifically focus our investigation on image spatial resolution with a cylindrical scanner design with the axial extent of the scanner increased to 2 m, and we quantify and compare the parallax blurring in the axial direction and the more commonly characterized radial parallax blurring in the transverse imaging plane. Here we only investigate hardware-based approaches, e.g. different detector configurations, to improve image resolution. It is known that the effects of parallax can be mitigated with a detector that measures DOI and we have simulated several scenarios of 2-layer and 3-layer detectors to study the impact on both the transverse and axial spatial resolution. We have performed these studies with a range of axial acceptance angles (up to the maximum of $\alpha = \pm 67^\circ$). This study will provide guidance for the choice of α for a particular task that may be affected by the tradeoff of sensitivity vs. spatial resolution. Equivalently, one can consider this a study of trade-offs in scanner designs with varying scanner length.

The outline of the paper is as follows: we begin with a description of the simulation methods used and the simulated experiments, we then characterize spatial resolution degradation in the radial direction and the improvement with DOI implementations for two different scintillation materials, and lastly we evaluate the effect of axial parallax error as the axial acceptance angle is increased from $\pm 12^\circ$ to $\pm 67^\circ$.

2. Methods

2.1 Simulated scanner geometry

We performed computer simulations using an EGS4-based Monte Carlo simulation package (Surti *et al* 2004). Annihilation events were generated without effects from positron range and photon non-collinearity and only true coincidence events were recorded for this investigation. It is important to note that there will be a significant increase in photon non-collinearity blurring for very oblique axial lines-of-response, as there is an ~84 cm detector separation for coincidences occurring within the same axial ring to ~217 cm for events occurring between the proximal ends of the scanner. This effect is not related with

geometrical parallax blurring and not correctable by DOI measurement, and therefore was not included to simplify interpretation of results. The PET scanner geometry was modeled as a cylinder of pixelated crystal rings with an axial length of 198 cm (see figure 1). The ring diameter was fixed at 85 cm. Two different scintillation materials were studied, LSO (also representative of LYSO, used in many commercial TOF PET/CT scanners) and LaBr_3 . We chose these two materials since they span a range of stopping power in scintillation crystals available today with TOF capability, and therefore represent the range of spatial resolution effects due to parallax. For LSO the attenuation length $\lambda = 1.2$ cm at 511 keV, while for LaBr_3 $\lambda = 2.1$ cm. To compare only the effect of scintillator stopping power, and not differences in crystal length, we kept the crystal size fixed at $4 \times 4 \times 20$ mm³ and the crystal pitch fixed at 4.3 mm, which is representative of detector designs using powder-based reflector material between individual crystals as in the *La-PET* scanner (Daube-Witherspoon *et al* 2010). Several examples of discrete DOI encoding schemes were used in the simulation, modeling stacked crystal designs that have been shown to be feasible to put into practice, for example determining the layer of interaction by pulse shape discrimination (Ito *et al* 2011). For 2-layer DOI detectors both equal layer thickness (10 / 10 mm) and unequal layer thickness (7 / 13 mm) DOI configurations were used (see figure 2). The rationale for choosing a design with unequal layer thicknesses aimed to have a similar number of events in each layer; for events directed perpendicular to the detector surface the sensitivity in each crystal layer is ~50/50% for a 7 / 13 mm combination of layer thicknesses, and ~65%/35% for equal layer thicknesses (10 / 10 mm). This relationship between layer thickness and layer sensitivity is strongly dependent on the angle of the incident photon; for oblique events with an incidence angle of 60°, the 7 / 13 mm combination of layer thicknesses will have a layer sensitivity of 65%/35% respectively, and the equal layer thickness will have a layer sensitivity of 80%/20%. The 3-layer DOI configuration used approximately equal layer thicknesses (6.5 / 6.5 / 7 mm). Positioning of events within a discrete crystal volume was performed using a center-of-gravity calculation when energy was deposited in multiple crystal elements. This same methodology was extended when assigning a discrete DOI bin if inter-crystal scattering occurred depositing energy in multiple crystal layers. Additional DOI blurring due to the detector's limited DOI resolution were not included. A lower energy threshold of 440 keV was applied in all simulations.

2.2 Spatial resolution phantom

The spatial resolution phantom was composed of seven point sources embedded within a warm background of uniform activity (see figure 3)—the warm background was included because an iterative reconstruction algorithm with a non-negativity constraint was used. See details in section 2.3. The phantom had a diameter of 50 cm and a thickness of 20 cm. The simulated phantom was composed of air (both the warm background and the point sources) to eliminate effects of object attenuation and object scatter. The seven point sources had a diameter of 0.5 mm and were positioned at radial offsets of 0 cm, 3 cm, 6 cm, 9 cm, 13 cm, 18 cm, 23 cm. The sources were located within a transverse plane 1 cm offset from the center of the axial FOV, to avoid the exact center of the scanner where the sampling density of lines of response may be very high. The activity ratio used was 25:1 (point source to background). Due to the large partial volume effects present using a detector with 4 mm pixels, the measured peak to background in the reconstructed images was much lower, and

we found 25:1 to be adequate for proper fitting and characterization of the point source response. These image spatial resolution measurements are not meant to be absolute, as compared with NEMA standard measurements, but relative (and representative) measures of spatial resolution.

2.3 Image reconstruction

Images were reconstructed using blob-basis functions (Matej and Lewitt 1996) and a line-of-response (LOR) based list-mode OSEM algorithm (Popescu 2004). We chose to use an iterative reconstruction algorithm rather than a linear analytic reconstruction method (e.g. filtered back-projection) because a LOR based approach is easier to implement for the data from a long-axial FOV system—in addition to being the most common method used clinically. Analytic reconstruction methods typically require rebinning and interpolation of axial projection data that can cause blurring in the axial direction for relatively short scanner axial FOVs; this effect will be more pronounced with a longer AFOV having a larger maximum acceptance angle. Our study therefore did not include analysis of data from analytic reconstruction methods where non-intrinsic effects, like the rebinning and interpolation of axial data, would confound the interpretation of results. In section 3.1 we show that measured transverse resolutions using OSEM are comparable to results using the analytic 3D-FRP algorithm (Matej and Lewitt 2001) reported in a previous publication, demonstrating that the imaging parameters used in this work—the point source to background activity ratio and stop iteration—provide viable spatial resolution results. Note that incorporation of the blob-basis functions in our OSEM algorithm provides an effective means to optimize the image signal-to-noise with good spatial resolution, compared to more conventional methods of regularization.

The OSEM reconstruction used 25 subsets and convergence was reached at iteration 4; this stop iteration was used for all image quantification. The position of each LOR is determined by the physical location of the center of the crystal. When DOI was used for reconstruction the radial position of the LOR was determined by the center of the crystal layer in which the interaction occurred—for DOI configurations having two layers of unequal length (7 cm and 13 cm), the LOR for interactions in the second layer originates 13.5 cm from the front face of the first crystal layer. The reconstructed image is generated with 2 mm cubic voxels. No point spread function (PSF) modeling was used in the image reconstruction. LORs between all axial ring combinations were used for the maximum acceptance angle $\alpha = \pm 67^\circ$; α was also restricted to $\pm 58^\circ$ (corresponding to an axial FOV of 137 cm), $\pm 40^\circ$ (corresponding to an axial FOV of 72 cm), and $\pm 12^\circ$ (corresponding to an axial FOV of 18 cm). Because the spatial resolution phantom used in this work was simulated as being composed of air, no attenuation or scatter corrections were applied.

2.4 Image analysis

To quantify the image spatial resolution we used the procedure described below. First, a square ROI having a 10-pixel width was used to estimate the count level in the image background. This background was then subtracted from the image. For each point source, line profiles were calculated to quantify the resolution in all three dimensions. The line profiles had a width of 3 pixels, to reduce noise and variability. A Gaussian fit was then

applied to the line profile data to determine the peak value and distribution centroid. The value of the FWHM and FWTM was calculated using a linear interpolation between pixels. FWHM calculated using linear interpolation, rather than using the Gaussian fitting, was determined to be a better metric since the shape of the response function was found to have longer tails than the Gaussian shape for the large axial acceptance angle. Transverse resolution is defined as the average FWHM from two orthogonal directions in the transverse plane (x and y directions) and axial resolution is defined as the FWHM from the line profile in the axial direction. The simulation output was divided into three independent list files, each reconstructed separately, for analysis of error. Error bars in figures show ± 1 standard deviation.

3. Results

3.1 Radial parallax error from point source simulations

Scanner spatial resolution as a function of position radially offset from the center of the FOV is characterized in figure 4 for LSO and LaBr₃ based scanners having an axial acceptance angle $\alpha \pm 40^\circ$. The transverse spatial resolution does not vary with axial acceptance angle at a radial position of 0. Compared to the LSO scanner, the transverse resolution with LaBr₃ is slightly worse due to the lower stopping power and higher probability of inter-crystal scattering. The measured FWHM of point sources from LSO scanners was approximately 4.9 mm, and the point source FWHM of LaBr₃ scanners measured 5.3 mm—these results are very similar (4.8 mm with LSO and 5.2 mm with LaBr₃) to previous simulations using the same scanner ring diameter, crystal size, and crystal pitch but reconstructed using the analytic 3D-FRP algorithm with a ramp filter and without a uniform background activity (Surti *et al* 2013). Both the LSO and LaBr₃ scanners showed degraded spatial resolution as the radial offset was increased, due to the inherent radial parallax error which blurred positioning in the transverse directions; the point source with the highest radial offset (23 cm) was chosen to represent a lesion at the edge of a very large patient. Incorporating a two-level DOI correction improved the transverse resolution at radial positions greater than 9 cm, however the measured resolutions for DOI scanners remained slightly degraded compared to measurements at the center of the imaging FOV having no radial offset. A detector having three-layer DOI encoding further improved spatial resolution at all radial offsets, though the impact was less compared to the improvement observed between no DOI and two-layer DOI. The further improvement in resolution with a three-layer DOI detector was more pronounced with LSO compared with LaBr₃, suggesting that DOI mispositioning caused by inter-crystal Compton scattering is limiting additional recovery of resolution.

For two-layer DOI implementations with different layer thicknesses, there was no difference in spatial resolution between equal length (10 / 10 mm) DOI layers and unequal DOI layers that have a shorter top layer and longer bottom layer (7 / 13 mm); a similar result was obtained in (Zhang *et al* 2013). We believe this is because a large fraction of coincidence events occurs at different depths in each DOI detector. For the unequal length DOI implementation (7 / 13 mm), the volumetric response in the LOR-based reconstruction algorithm for events occurring in top layer of one detector (the 7 mm layer) and the second event in the bottom layer (the 13 mm) is roughly the same as LORs with the equal length

(10 / 10 mm). An unequal (7 / 13 mm) 2-layer DOI detector may be advantageous for normalization correction techniques, as the efficiency of each layer is roughly equal, and this will be of heightened importance in a long-axial FOV system with many LORs.

3.2 Parallax error in scanners with different axial FOV

The transverse resolution at the center of the imaging FOV (no radial offset) for different axial acceptance angles is shown in figure 5. As observed in results shown in figure 4, there are systematic differences in spatial resolution between LaBr_3 and LSO scanners, corresponding to the differences in the scintillator stopping power. There is a slight improvement in transverse resolution at the center of the FOV with DOI, presumably due to an increase in sampling. Only the 2-layer equal-thickness (10 / 10 mm) DOI configuration is shown in figure 5. Shown in figure 6 is the point source FWHM in the axial direction for scanners of different axial extents, with the point source placed in the axial and transverse center of the scanner. The point source FWTM for scanners having different axial extents is shown in figure 7. The FWTM increases at a greater rate, compared to the FWHM, as the axial acceptance angle increases. The increased FWTM was more pronounced in the LaBr_3 scanner designs, where the lower stopping power and increased inter-crystal scatter also contributed to a larger FWTM.

The results show a degradation in axial resolution when extending the axial acceptance angle from $\alpha = \pm 12^\circ$ to $\pm 58^\circ$. There was little change in axial resolution when extending $\alpha > \pm 58^\circ$. This is due to the point source emission geometry, and the relatively small increase in solid angle coverage when increasing α beyond this value, which adds only a small fraction of coincidence events recorded between the proximal ends of the scanner (the number of recorded events increased by 40% when increasing from $\alpha = \pm 40^\circ$ to $\pm 67^\circ$ and only 11% when increasing from $\alpha = \pm 58^\circ$ to $\pm 67^\circ$). In figure 8, the number of recorded coincidence events in the axial direction along a row of crystals is shown as a function of axial acceptance angle to illustrate the diminishing return in sensitivity as the acceptance angle increases. Also shown in figure 8 is the impact of object attenuation with an average sized distribution—here we assume a 30 cm diameter water filled cylinder, 2 m long. It is important to note that in the presence of object attenuation there will be relatively fewer oblique LORs, and therefore, the degradation of axial resolution as α increases will be reduced compared to the results shown in Figures 6 and 7.

4. Discussion

Here we present results from a Monte Carlo simulation study specifically designed to characterize the effect of parallax error in the axial dimension of reconstructed images as the acceptance angle α of a 198 cm long PET scanner is increased from $\alpha = \pm 12^\circ$ (typical for a scanner with 18 cm length) to the maximum of $\pm 67^\circ$. Simulated images of point sources at the center of the axial and transverse FOV (where axial parallax is highest) and at increasing radial offsets (where radial parallax dominates) aim to characterize the magnitude of radial and axial parallax blurring and their contributions to loss of spatial resolution as a function of position in the imaging FOV.

One of the motivations behind designing a PET scanner with a very long axial extent is to improve the intrinsic sensitivity by increasing the solid angle coverage of the scanner, however these sensitivity gains are only possible if the axial acceptance angle is enlarged to include these additional LORs. As the axial acceptance angle is increased, more axial oblique LORs will be included, and it is these LORs that will cause blurring in the axial direction of reconstructed images. It should also be noted that a primary intention of a long axial FOV scanner is to image the entire body simultaneously, in which case the maximum degradation of spatial resolution will only pertain to structures in the center of the FOV assuming the axial acceptance angle is unrestricted. With modern, fully-3D iterative reconstruction algorithms it is not straightforward to predict the magnitude of this resolution degradation, and the impact of DOI correction, and therefore simulation has proven to be a useful tool in the optimization of the scanner design and choice of data acquisition parameters for a long-axial FOV system. The experimental geometry that was simulated was chosen to emphasize the effect caused by axial parallax error: the point sources are positioned at the center of the transverse and axial FOV, there is no object attenuation which would reduce the collection of events at very oblique axial angles, and there was no object scatter which would also cause a reduction in contrast. We have learned in this study that the maximum degradation in axial spatial resolution is ~19% (for a 198 cm LSO scanner) compared to an 18 cm long scanner which is a typical axial length of commercial PET/CT systems today. There was no increased degradation in transverse resolution with increasing axial acceptance angle, and therefore the change volumetric resolution was also ~19% for $\alpha = \pm 67^\circ$. This modest loss in spatial resolution can be compared to the very significant gain in sensitivity for the 2 m long scanner of ~40 $\times$ for the case of total body imaging (Badawi *et al* 2013). Alternatively, we can consider how to optimize the sensitivity gain for the 2 m long scanner by restricting the axial acceptance angle. Using a simple distribution of a line source along the central axis, following the NEMA definition of intrinsic scanner sensitivity, we can easily estimate this effect. With an unrestricted axial acceptance angle the sensitivity is ~5.8 $\times$ higher than if we set $\alpha = \pm 12^\circ$ to correspond with an axial FOV of 18 cm in order to ensure that the axial resolution does not degrade beyond 4.7 mm. It is also possible to optimize α , to find a more suitable trade-off of sensitivity and spatial resolution, for a specific imaging task.

We also compare axial parallax blurring with radial parallax blurring to put these results into perspective with the relative losses in spatial resolution due to the geometry of a ring system. Point sources were simulated radially offset from the center of the FOV to measure this effect. The transverse FWHM (average FWHM along x and y directions) of a LSO-based scanner with $\alpha = \pm 40^\circ$ degraded from 4.9 mm, at the radial center, to 6.5 mm at a radial offset of 18 cm and 7.7 mm at a radial offset of 23 cm. It is difficult to directly compare the effects of parallax in the transverse direction to the axial direction since the relative fraction of oblique LORs suffering from parallax impacts the 3D image reconstruction in a different way. In addition, we are only sampling a small sub-set of locations in the FOV to emphasize these effects—points at the center of the FOV emphasize the degradation in axial resolution and points at large radii emphasize the degradation in transverse resolution—however for the locations studied in this work radial parallax shows a much stronger influence on resolution than axial parallax.

To consider the effects of parallax on a realistic imaging task requires a more involved simulation, such as the one performed by Surti and Karp 2015, which considered lesion contrast and detectability for a scanner with a 72 cm long axial FOV. Note that we studied $\alpha = \pm 40^\circ$ in this current study to correspond with a scanner length of 72 cm. We can understand the previous results in terms of the findings in this current study of spatial resolution. Our simulation model of a two-layer DOI detector (with modeling of inter-crystal scatter included) shows some improvement in the resolution, in both the response at FWHM and FWTM for scanner designs based on both scintillation materials (see figures 4–7), however, this improvement may not be enough to significantly affect the image quality metrics studied in Surti and Karp 2015. Note, in that study the 1-cm diameter lesions were placed in the center of the axial FOV at radial offsets of 7 cm and 13 cm, and does include attenuation and scatter. This was designed to be representative of quantitative imaging tasks in oncology and cannot be generalized to all applications, but, together with this focused study on spatial resolution, they suggest that scanners with very long axial FOV can be built for whole-body imaging using current TOF detectors with no DOI capability without causing large reductions to image quality. Nevertheless, a practical and simple method to include DOI capability into the detector design would obviously be an advantage for all PET system designs, long and short AFOV alike. An investigation into optimal image reconstruction approaches may further improve the imaging capabilities of a 2 m long scanner. Incorporating an accurate model of the scanner resolution properties is likely to further reduce degradations caused by parallax error and improve spatial resolution.

5. Conclusions

In this paper image spatial resolution was studied for long-axial FOV PET system designs, specifically focusing on the effect of axial parallax error. For long-axial FOV PET system designs additional gains in sensitivity can be achieved by increasing the acceptance angle of axial LORs, however these LORs have a greater parallax in the axial direction and therefore sensitivity gains should be assessed with degradations to image spatial resolution. Simulations of a cylindrical PET scanner having an axial length of 2 m and a 85 cm ring diameter with $4 \times 4 \times 20 \text{ mm}^3$ LSO crystals showed that the axial FWHM point source response for sources in the center of the FOV degraded from 4.8 mm to 5.7 mm as the axial acceptance angle increased from $\alpha = \pm 12^\circ$, typical for a scanner with 18 cm length, to the maximum of $\pm 67^\circ$. For a system based on LaBr_3 scintillator the axial FWHM point source response showed a similar degradation of $\sim 1 \text{ mm}$ for the same conditions. There was no significant change in transverse resolution with increasing axial acceptance angle for both scintillation materials studied. A two-layer DOI detector improved both axial and radial resolution, however the results show a small change in volumetric resolution caused by the use of a large axial acceptance angle, and therefore improvements observed using a DOI detector in a long-axial FOV system are similar to observations made with standard length PET scanners ($\sim 20 \text{ cm}$)—the predominant benefit of DOI correction is to improve radial parallax error. We conclude that for a fully-3D, 2 m long scanner the impact of axial parallax is small when using modern iterative image reconstruction methods, and therefore the use of current, non-DOI detectors can be used in long-axial FOV design without significant degradations to image resolution.

Acknowledgments

The authors would also like to thank Dr Ramsey Badawi and Dr Simon Cherry for valuable discussions. This work was supported by the National Cancer Institute of the National Institutes of Health under award numbers: R01CA113941 and R01CA170874.

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Long-axial FOV scanner

Figure 1.

Illustration of a 2 m long scanner design simulated in this work showing the scanner's axial extent and maximum axial acceptance angle. The scanner design had an 85 cm ring diameter.

Figure 2.

Depth encoding configurations that are simulated in this work. Layer thicknesses used: 7 mm top 13 mm bottom, 10 mm top 10 mm bottom (equal thickness), and a three layer detector with approximately equal lengths.

Figure 3.

Reconstructed images of the spatial resolution phantom composed of seven point sources in a warm background that was used in this work. The point sources are positioned at radial offsets: 0 cm, 3 cm, 6 cm, 9 cm, 13 cm, 18 cm, 23 cm. The activity ratio was 25:1 (point source to background).

Figure 4. Calculated transverse spatial resolution for point sources with varying radial offsets in the center of the axial FOV. Several different 2-layer DOI encoding schemes were simulated and also a 3-layer (with equal lengths). The scanner axial acceptance angle was fixed at $\alpha = \pm 40^\circ$ and the crystal size was $4 \times 4 \times 20 \text{ mm}^3$ for both LSO and LaBr₃.

Figure 5. Spatial resolution in the transverse plane ($r=0$) at the center of the axial FOV for different axial acceptance angles of $\alpha = \pm 12^\circ, \pm 40^\circ, \pm 58^\circ, \pm 67^\circ$ (corresponding to axial extent of 18cm, 72cm, 137cm, 198cm). A 2-layer DOI detector having equal layer thicknesses was used.

Figure 6. Spatial resolution along the axial direction for different axial acceptance angles of $\alpha = \pm 12^\circ$, $\pm 40^\circ$, $\pm 58^\circ$, $\pm 67^\circ$ (corresponding to axial extent of 18 cm, 72 cm, 137 cm, 198 cm). A single point source in the center of the axial FOV was used. A 2-layer DOI detector having equal layer thicknesses was used.

Figure 7.
Calculated FWTM from point source measurements for different axial acceptance angles (as in figure 6). A 2-layer DOI detector having equal layer thicknesses was used.

Figure 8.

Recorded coincidence events for a single row of crystals along the scanner's axial extent (plotted as a function of the LOR angle). A single point source at the center of the imaging FOV, with no uniform background added, was used. Here we show the effect of attenuation on the distribution of collected events assuming the point source is in the center of a water filled cylinder, 30 cm in diameter and 2 m long. The counts are normalized to the angular bin with the largest number of counts.