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Abstract. Positron emission tomography (PET) scanners designed for imaging of small animals have transformed translational research by reducing the necessity to invasively monitor physiology and disease progression. Virtually all of these scanners are based on the use of pixelated detector modules arranged in rings. This design, while generally successful, has some limitations. Specifically, use of discrete detector modules to construct PET scanners reduces detection sensitivity and can introduce artifacts in reconstructed images, requiring the use of correction methods. To address these challenges, and facilitate measurement of photon depth-of-interaction in the detector, we investigated a small animal PET scanner (called AnnPET) based on a monolithic annulus of scintillator. The scanner was created by placing 12 flat facets around the outer surface of the scintillator to accommodate placement of silicon photomultiplier arrays. Its performance characteristics were explored using Monte Carlo simulations and sections of the NEMA NU4-2008 protocol. Results from this study revealed that AnnPET's reconstructed spatial resolution is predicted to be ~ 1 mm full width at half maximum in the radial, tangential, and axial directions. Peak detection sensitivity is predicted to be 10.1%. Images of simulated phantoms (mini-hot rod and mouse whole body) yielded promising results, indicating the potential of this system for enhancing PET imaging of small animals. © 2017 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JMI.4.1.011007]

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1 Introduction

Typically, small animals, mainly rodents, are used in biomedicine as models for disease. Animal dissection was routinely utilized to examine the model's progression and/or its effect on physiology and anatomy at numerous time points, requiring large numbers of animals. Application of this paradigm has been reduced over the last two decades with the introduction of high-resolution scanners designed for imaging of small animals.^{1–8} It is now possible to perform longitudinal studies on relatively small cohorts of animals. The application of radiopharmaceutical-based molecular imaging techniques is of particular interest to investigators due to the ability to track and quantify physiologic processes.

Since introduction of the first positron emission tomography (PET) scanners designed for use with small animals in the 1990s, their spatial resolution and detection sensitivity have improved.^{3,4,9–37} To achieve high resolution, these scanners utilize multiple detector modules consisting of arrays of small, discrete detector elements arranged in rings. This design, however, creates regions where there is little or no scintillator at the joints between modules and at the spaces between detector elements.³⁸ The presence of these gaps results in reduced detection sensitivity and, in some cases, image artifacts.^{38,39} St. James et al.³⁸ reported a 64% increase in the sensitivity of a PET scanner constructed with tapered scintillator elements (reducing, but

not eliminating the gaps between modules) compared to a PET scanner utilizing straight detector elements where relatively large gaps are present. Furthermore, construction of arrays of very small scintillator elements necessary to produce the high spatial resolution required for preclinical PET imaging is labor intensive, expensive, and complex (a limited number of commercial vendors are capable of delivering these arrays).

To address some of the challenges associated with the creation of preclinical PET scanners based on arrays of discrete detector elements, some researchers have investigated modules using a monolithic scintillator.^{36,40,41} In these systems, detection sensitivity is improved by eliminating the nonscintillating regions between adjacent detector elements present in pixelated detectors. These regions can represent nontrivial fractions of the area of a detector, especially when very small elements are used. A unique capability of monolithic scintillator-based detectors is the ability to correlate event depth-of-interaction (DOI) in the scintillator with the shape of the light distribution impinging upon the photodetectors.^{42–45} Thus, it is possible to correct data for the parallax effect.³⁶ Previous studies utilizing flat pieces of continuous scintillator reported accurate measurements of DOI at the center of the detector, but degraded performance at the edges of the detector caused by distortions of the light distribution.^{42–45} Since these scanners arrange the continuous detector modules in a ring, gaps exist where the modules meet (as in the use of pixelated modules), resulting in a loss of detection sensitivity.

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To address the continuing challenges in applying monolithic scintillator-based detectors to the construction of high resolution and efficient preclinical PET scanners, we are exploring a system based on a monolithic annulus of lutetium-yttrium oxyorthosilicate (LYSO) coupled to arrays of silicon photomultipliers (SiPM), called AnnPET. Two other groups have investigated the use of an annular scintillator. Genna and Smith constructed the ASPECT scanner in 1988 from an annulus of NaI(Tl).⁴⁶ Freifelder et al.⁴⁷ constructed the HEAD PENN-PET scanner, which utilized an annulus of NaI(Tl) (inner diameter = 42 cm, overall diameter = 80 cm, and axial length = 30 cm) connected to PMTs. Other than the use of NaI(Tl), HEAD PENN-PET differs from our system in that it was designed for the imaging of human heads. Therefore, by necessity, its bore size was 8.5 times larger and 2.4 times shorter than AnnPET. Furthermore, use of SiPMs makes the new system compact and virtually immune to the effects of magnetic fields, making it possible for it to be used in conjunction with an MRI scanner. Finally, in contrast to HEAD PENN-PET, AnnPET takes advantage of the use of a continuous scintillator to measure DOI. Since our scanner uses a single piece of scintillator, it does not require construction of expensive pixelated scintillator arrays or support structures required to create conventional PET scanners. This system retains the positive qualities of scanners based on a monolithic scintillator (elimination of dead spaces between detector elements and DOI measurement capabilities), while eliminating gaps between detector modules. This work describes an initial exploration of the basic design, photon transport characteristics, and performance of the AnnPET scanner performed with computer simulations.

2 Methods

2.1 Detector Geometry Definition

The simulated AnnPET scanner consisted of a 7.2-cm long annulus of LYSO with an outer diameter of 8 cm and inner diameter of 5 cm. This size is the longest and widest uniform boule of LYSO currently available from commercial scintillator vendors. Twelve 1.9 cm × 7.2 cm × facets were placed equidistantly around the outer surface of the annulus to permit attachment of SiPM arrays. Note that it is possible to create this scanner with existing methods at lower cost compared to equivalent scanners due mainly to reduced detector construction labor

Fig. 1 Schematic drawing of the AnnPET scanner showing the scintillator and SiPM arrays.

costs (approximately a factor of two lower). The distance between opposing SiPMs is 7.5 mm (Fig. 1). Simulation of the detector was performed using the Geant4 Application for Tomographic Emission (GATE) software package running on a multicore (32 core) computing cluster. GATE is an advanced open source software package developed by the international OpenGATE collaboration to emulate the performance of PET and SPECT scanners. It has been extensively validated and used in numerous projects to evaluate new PET scanner designs.^{48–63} Without simulations, it is very challenging to explore new scanner designs, due to the cost and complexity of creating prototype systems intended solely for design evaluation purposes. AnnPET was modeled using the C-PET GATE scanner definition with a 6 ns coincidence timing window and 15% energy resolution; positron transport, Compton scattering, and photoelectric interactions were included in the model.

2.2 Photon Transport Model

Acquisition of data from the AnnPET scanner required simulations of interactions of annihilation photons with the scintillator, and then transport of the optical photons resulting from these interactions to the SiPM arrays mounted on the outer surface of the scintillator. A flowchart illustrating this process is shown in Fig. 2. As noted above, simulation of positron emissions, their annihilations, and photon interactions (photoelectric effect and Compton scattering) in the scintillator were performed with GATE. Transport of scintillation photons to the SiPM arrays was modeled using the DETECT2000 software environment.⁶⁴ DETECT2000 is a Monte Carlo software environment for simulating the behavior of optical systems, and has been used to study the characteristics of PET detectors.^{65,66} For example, Miyaoka et al.⁶⁷ used DETECT2000 to model continuous-scintillator detectors to aid in improving event position determination. To reduce reflection of photons into the detector volume

Fig. 2 Flowchart describing the multistep process for producing AnnPET data.

from the two end caps (an effect that could distort the photon distributions at the ends of the detector), these surfaces were blackened. This technique has been used by other groups to reduce this effect in detectors based on continuous slabs of scintillator.^{42–45} It should be noted that unlike PET scanners based on arrays of continuous-scintillator detector modules, AnnPET will only experience edge effects at the two ends of the scanner instead at every interface between detector modules. Finally, the inner surface of the annulus was coated with D’Lambertian reflector to diffusely reflect scintillation light.

Unfortunately, accurate modeling of photon transport is computationally intensive and hence, time-consuming. To make the emulation of the photon transport effects more efficient (enabling the simulation of complex phantoms), we developed a method to rapidly estimate these effects. The outline of this process is shown on the left side of Fig. 2 (outlined in red). The first step was the creation of representative annihilation photon interaction points in the scintillator. A simulated ²²Na point source at the center of the detector was scanned, resulting in 200,000 total events. The three-dimensional (3-D) positions of the annihilation interaction points created by the GATE simulation were used by the DETECT2000 software as scintillation photon initiation points. The number of photons created at each point was proportional to the amount of energy deposited in the scintillator (32,000 photons/MeV⁶⁸). DETECT2000 was then used to track the photons in the scintillator, through a simulated 0.2-mm layer of optical coupling gel and finally to the simulated photodetector surfaces that possess the absorption properties of SensL C-series SiPM arrays.⁶⁹

The result of the DETECT2000 simulation was a map of the distribution of the photons detected by the SiPM arrays for each annihilation photon interaction in the scintillator. In order to estimate coordinates of the origin of the emitted photons, a rotational transformation was first applied. The rotation established a coordinate system in which the SiPM surfaces were parallel to the $x - y$ plane and event penetration depths in the scintillator were parallel to the z -axis. The x - and y -coordinates were then histogrammed into bins representing the physical dimensions of the SensL C-Series SiPM arrays (5×19 array of 3×3 mm² SiPMs, pitch = 4.2 mm). The binned distributions were fit with Gaussian functions. Data included in the fitting procedure were confined to photons within one detector facet to either side of the maximum histogram bin. Annihilation events that produced non-Gaussian photon distributions at the surface of the SiPM arrays (caused mainly by scattering of photons at the inner surface of the annulus or absorption at the ends of the scanner) were discarded and not used in subsequent analyses. Centroids of the fits were defined as x - and y -coordinates of the photons’ origins. Z -coordinates (DOI) were first estimated by taking a ratio of the total number of counts in the photon distribution to its peak intensity (N/I).⁷⁰ Coordinates were then calculated using the methodology described by Lerche et al.⁴² and successfully applied by Gonzalez et al.⁷¹ that fits the photon distribution to a function that models photon attenuation and transport in the scintillator. The estimated event coordinates were compared to the known coordinates to determine the errors in position calculations for each event (Δ_x , Δ_y , and Δ_z). The uncertainties from all of the events were combined to produce average errors in the three coordinates, which were used to create probability density functions modeling the process of measuring event position (x - and y -coordinates) and depth (z -coordinate):

$$f(x|x_0, \bar{\Delta}_x^2) = e^{-\frac{1}{2}\left(\frac{x-x_0}{\bar{\Delta}_x}\right)^2}, \quad (1)$$

$$f(y|y_0, \bar{\Delta}_y^2) = e^{-\frac{1}{2}\left(\frac{y-y_0}{\bar{\Delta}_y}\right)^2}, \quad (2)$$

$$f(z|z_0, \bar{\Delta}_z^2) = e^{-\frac{1}{2}\left(\frac{z-z_0}{\bar{\Delta}_z}\right)^2}. \quad (3)$$

The parameters x_0 , y_0 , and z_0 are the actual coordinates of the annihilation photon interactions in the scintillator; x , y , and z are the coordinates of the events as determined by the scanner following modeling of photon transport to the SiPMs; $\bar{\Delta}_x$ (1.9 mm), $\bar{\Delta}_y$ (1.9 mm), and $\bar{\Delta}_z$ (6.0 mm) are the average uncertainties in the determination of x , y , and z . The uncertainties are due mostly to the effect of scatter from the surfaces of the annulus, resulting in a broadening and shifting of the optical photon distributions. These functions were then used in subsequent simulations to estimate the x -, y -, and z -coordinates of annihilation photon interactions in the scintillator.

2.3 Scanning Process

The simulated scanning process is shown on the right side of Fig. 2 (outlined in blue). The first step is the definition of the scanner geometry in GATE (described above). Next, the physical characteristics (size, shape, and composition) and radionuclide distribution (¹⁸F or ²²Na) of the simulated object are specified. The simulation is then initiated. The end result of the simulation process is a list mode file. Each line contains the 3-D coordinates and energy deposited in the scintillator for each of the coincident annihilation photons. The coordinates specify the actual locations of the events in the scintillator, but not the ones that would be recorded by the actual AnnPET scanner. This determination requires introduction of the errors inherent in position calculations based on the distribution of scintillation light impinging on the SiPM arrays. To achieve this goal, the probability density functions given by Eqs. (1)–(3) were sampled. Hence, the original GATE-produced list mode file containing the actual positions of the events was transformed into one that incorporates the effects of positioning errors and DOI measurements. A set of sinograms, spanning the entire detector field-of-view, was then created from the transformed list mode data. Images were reconstructed with either the single slice rebinned-filtered backprojection (SSRB-FBP) algorithm for NEMA-based measurement of spatial resolution or with our MLEM reconstruction software.⁷²

2.4 Performance Measurements

Basic performance characteristics of AnnPET were assessed using NEMA NU4-2008-based tests.⁷³ Spatial resolution was measured by simulating a point source of ²²Na (50 μ Ci) mounted in a 1 cm³ block of acrylic positioned at several radial positions at the axial center of the scanner. The full widths at half maximum (FWHMs) of intensity profiles acquired from SSRB-FBP images of the point source were reported as spatial resolution. Detection sensitivity was measured by simulating the same point source at nine positions along the central axis of the scanner.

High resolution imaging capabilities of AnnPET were illustrated by simulating a mini-hot rod phantom consisting of four,

4×4 arrays of rods filled with ^{18}F embedded in a tank of water. The diameters of the rods were 0.75, 1.0, 1.25, and 1.5 mm (corresponding center-to-center distances were 1.5, 2.0, 2.5, and 3.0 mm). Finally, to observe the potential performance of AnnPET applied to imaging of a small animal, the brain of the 4D mouse whole body (MOBY) digital phantom (v2.0)⁷⁴ was used with the simulated scanner. The amount of activity in each of the phantom's structures was adjusted to simulate a 100 μCi injection of ^{18}F -fluorodeoxyglucose.⁷⁵ The simulated phantom was scanned for 10 min. Since the ranges of positrons, even those emitted by ^{18}F (mean and maximum range of ~ 0.6 and ~ 2.4 mm, respectively^{76,77}), are equivalent to the predicted scanner resolution, it was necessary to correct the images for this effect. Consequently, a modified version of Derenzo's deconvolution method,⁷⁸ using a 3-D kernel based on the positron range distribution for ^{18}F was developed⁷⁹ and applied to the images of the hot rod and MOBY phantoms.

3 Results

Events in all areas of the scintillator (including those located at the seams between two facets) were identifiable. Figure 3 shows a representative detected photon distribution from an event located at the center of the scanner (a) and ~ 4 mm from the top end cap (b). The widths of the detected photon

distribution (FWHM) are plotted in Fig. 4 as a function of annihilation photon interaction point distance from the SiPM surface (a) and axial position of the event (b). Figure 5 shows the plots of the ratio of the mean number of photons detected by the SiPMs used to calculate event position to the total number of photons detected by all of the SiPMs in the scanner (also shown are the minimum and maximum ratios) as a function of annihilation photon interaction point distance from the SiPM surface (a) and axial position of the event (b). These ratios indicate that $\sim 83\%$ of the photons emitted in the scintillator is detected by the SiPMs used to calculate position (range 62% to 96%). Thus, only about 17% (range 38% to 4%) of the emitted photons scatter to distant areas of the scanner and are not used to calculate event position. The resolution of our DOI estimations (mean difference between actual depth and estimated event depth) plotted as a function of event depth is shown in Fig. 6. Figure 7 shows the plots of spatial resolution as a function of radial position of the source. Figure 8 shows a plot of detection sensitivity as a function of axial position. Peak sensitivity is 10.1%, which is considerably higher than most other scanners of comparable dimensions. A positron range-corrected image of the mini-hot rod phantom is shown in Fig. 9; intensity profiles drawn on the images of the rods are shown in Fig. 10. Finally, Fig. 11

Fig. 3 Plots of detected photon distributions at (a) the center of the detector and (b) at the top edge of the detector.

Fig. 4 Plots showing the mean FWHM of the photon distributions striking the SiPM arrays as (a) a function of event position from the SiPM array face and (b) a function of axial position.

Fig. 5 Plot of the mean ratio of photons detected to photons emitted as (a) a function of event position from the SiPM array face and (b) a function of axial position. Also shown are the minimum and maximum ratios.

Fig. 6 Plot of DOI resolution as a function of distance from the SiPM array surface.

shows a positron range-corrected image of the central region of the brain section of the MOBY phantom.

4 Discussion

As preclinical PET systems have emerged to become integral parts of research programs, ever-improving performance capabilities (spatial resolution and detection efficiency) are necessary to meet expanding applications. For example, the imaging of mice requires spatial resolutions of ~ 1 mm FWHM, or less, and high detection sensitivity to permit rapid scanning. To achieve these goals, we explored a new scanner design; one based on a single annulus of scintillator attached to arrays of SiPMs.

An important aspect of this new PET scanner design is use scintillation photon transport to gain information about event

Fig. 7 Plot showing spatial resolution (SSRB-FBP) as a function of radial position.

position in the detector. It is for this reason that we performed a detailed study of the characteristics of photon motion in the scintillator. Sample photon distributions detected by the simulated arrays of SiPMs are shown in Fig. 3. They possess the expected Gaussian shape. The distribution measured at the central area of the scanner [Fig. 3(a)] is symmetric. At regions closer to one of the two ends of the detector, however, the distribution becomes slight compressed along the axial dimension until clipping of the Gaussian occurs when the event is ~ 4 mm and less from the end [Fig. 3(b)]. The widths of these distributions plotted as a function of distance from the SiPMs [Fig. 4(a)] confirm that the width of the photon distribution is related to depth of the event (a phenomenon we plan to exploit to estimate DOI). The plot in Fig. 4(b) demonstrates that there is also a relationship between axial position of annihilation photon interaction point in the scintillator and shape of the detected photon

Fig. 8 Plot showing detection sensitivity as a function of axial position.

Fig. 9 Positron range-corrected images of the mini-hot rod phantom (rod diameters are shown).

distribution. This behavior is produced by the compression of the light distribution as the event approaches either end of the detector. Note that the blackened end caps reduce the number of reflected photons from these structures that could distort the Gaussian shape of the distribution. It is not until the event occurs close to an end cap that clipping of the distribution occurs [Fig. 3(b)]. Even at this position, fitting to a Gaussian is possible. It is not until the event approaches ~ 2 mm from the end cap that fitting of the photon distribution to a Gaussian function fails. Thus, recovery of event position in these regions is not possible. The dependence of width on axial position will necessitate a calibration procedure that correlates axial position with photon distribution size.

Unlike detector modules based on relatively small monolithic parallelepipeds of scintillator, it may be possible for scintillation photons to travel relatively long distances from an interaction point in the annulus, potentially making event

positioning difficult. Therefore, we studied the fraction of photons that were not detected by the SiPM arrays used in our position calculations. The results in Fig. 5 show that the photon distribution is relatively localized [a mean of 83% of all detected photons were localized to our detection area (± 1 detector facet)]. These plots also illustrate that this detection ratio is independent of event position. As described above, the shape of the detected photon distribution can be used to estimate the DOI for each event in the scintillator. The plot in Fig. 6 shows that the resolution of the DOI calculation depends upon the depth of the event. Specifically, DOI calculations for shallow events are slightly less accurate (mean DOI measurement resolution = 6.3 mm FWHM) than for deep events (mean DOI measurement resolution = 5.2 mm FWHM). This difference is due to uncertainties caused by the reduced number of photons emitted by shallow events reaching the SiPM due to photon attenuation in the scintillator.

As the plot in Fig. 7 shows, the spatial resolution of AnnPET in all three dimensions is ~ 1 mm FWHM (SSRB-FBP), which compares favorably to small animal PET scanners based on arrays of discrete detector elements whose resolution ranges from 1.63 to 2.32 mm (SSRB-FBP reconstruction).³⁷ The results also compare well with other small animal PET scanners based on monolithic scintillator whose reported spatial resolution range from 0.7 to 1.65 mm FWHM. Note that these results were obtained using iterative construction methods and, therefore, are not comparable to the results reported for AnnPET, which were determined using SSRB-FBP reconstructions as required by the NEMA NU4-2008 protocol. It is known that measurements of resolution utilizing iterative reconstructions of point sources in cold backgrounds can produce deceptively good results.⁸⁰ Some of the degradation of spatial resolution may be due to the use of average position uncertainties ($\bar{\Delta}_x$, $\bar{\Delta}_y$, and $\bar{\Delta}_z$) in the photon transport probability density functions [Eqs. (1)–(3)], which was necessary to make the modeling process more efficient. Finally, AnnPET's relatively good resolution uniformity is due to the modeling of event DOI measurement enabled by the use of a continuous scintillator.

One of the important potential advantages of utilizing a solid annulus of a scintillator is a lack of the gaps between detector modules and between detector elements (for pixelated detectors) present in most PET scanners. The absence of gaps minimizes dead areas in the detector, maximizing detection sensitivity. This effect is demonstrated in the plot of sensitivity shown in Fig. 8. Peak sensitivity (at the center of the scanner) is predicted to be 10.1%, which compares well with the peak sensitivities of scanners utilizing arrays of discrete detector elements (1.19% to 6.72%³⁷), and with those utilizing monolithic scintillator-based detectors (0.3% to 9%^{36,40,41}). Enhanced detection sensitivity increases the flexibility and capabilities available to researchers. For example, higher detection efficiency could be exploited to increase the temporal sampling of the radiotracer concentration in dynamic scanning studies by permitting the shortening of individual scan times, while maintaining sufficient counting statistics. Alternatively, the amount of radiotracer could be reduced without affecting count density in the resulting images. Also note that, since the only edges present in AnnPET are at the ends of the scanner, measurement of event position should not be significantly hampered by distortion of photon distributions caused by reflection/absorption at the edges of discrete, monolithic scintillator detectors noted by other investigators.^{42–45} Thus,

Fig. 10 Profiles drawn on the images of the hotrod phantom shown in Fig. 9. Each plot is labeled with the diameter of the respective rods.

Fig. 11 Image of the brain section of the MOBY phantom. (a) Slice from the MOBY phantom with major brain sections labeled [cortex (Cort), thalamus (Thal), hypothalamus (Hypo), caudoputamen (Caud), ventricles (Vent) (which contain no radioactivity), and amygdala (Amyg)], and (b) image of the phantom shown in (a). Note that the total diameter of the mouse brain phantom is ~ 1 cm.

imaging performance could be more uniform across AnnPET's field-of-view compared to a scanner using arrays of discrete, continuous scintillator modules that possess numerous boundaries between adjacent modules.

The image of the mini-hot rod phantom in Fig. 9 illustrates the potential advantages of high resolution imaging with AnnPET. Specifically, the 1.25- and 1.5-mm diameter rods are discernable from their neighbors, the 1.0-mm diameter

rods are marginally discernable, though it is not possible to identify individual 0.75-mm diameter rods. These visual impressions are supported by the intensity profiles drawn on the images (Fig. 10). The potential value of AnnPET is demonstrated by the image of the central area of the brain section of the MOBY phantom (Fig. 11), where it is possible to identify many of the small radiotracer-avid structures present in the mouse brain.

There are some engineering challenges to constructing the AnnPET system. First, performance of the SiPM arrays is enhanced by cooling.⁸¹ Normally, cooling would be accomplished with Peltier coolers, but the heat removed from the SiPMs by these devices must then be dissipated. Often this process is accomplished by circulating air or fluid over the Peltier coolers. While it is possible to place the necessary tubing to perform these tasks beside the scanner, the geometry of AnnPET may make this procedure complex. A potentially more efficient cooling method may be to immerse the SiPM arrays in cooled, electrically nonconductive fluid circulated through the sealed AnnPET gantry.⁸² In addition to specialized hardware, AnnPET may require modified software techniques. For example, identification of the position of individual events could use methods previously developed for monolithic scintillator detectors, similar to the ones described above. Specifically, the x - and y -coordinates can be calculated by determining the center-of-mass of the scintillation light distribution recorded by the SiPMs.⁶⁷ The DOI of events could be calculated by applying (N/I)-based modeling method used in this study. Note that nonuniformities of light output in the scintillator, which could affect the calculation of DOI, are normalized due to the use of the N/I ratio at the first stage of the depth calculation. Additionally, the effects of light production and collection variability in determination of event position (x - and y -coordinates) can be corrected by incorporating data acquired from a uniform flood phantom into the reconstruction algorithm.

To enhance the efficiency of the system, a flexible-zone-based data acquisition triggering system could be used to avoid pulse pile-up. Specifically, the outputs from individual SiPMs would not be grouped into rigidly defined zones (as in the HEAD PENN-PET scanner⁴⁷). Instead, the triggering system would identify the position of an event from any area of the scanner by identifying the maximum SiPM signal, and digitize only the SiPM outputs that are associated with this event (the size of the cluster of output to be digitized would be flexible and based on estimates of the number of SiPMs detecting light above a set threshold value). This type of sophisticated triggering system is required to produce accurate measures of energy deposition in the scintillator and estimate DOI. A flexible trigger is also necessary to enhance count rate capabilities of the efficient AnnPET detector (~10% peak detection efficiency) and process the events produced in the bulk scintillator material due to the presence of radioactive lutetium in LYSO. Seidel et al.⁸³ reported a count rate of 100 cps/cc caused by the presence of ¹⁷⁶Lu for a 200 keV energy window centered at 511 keV. For the annulus of LYSO tested in this investigation (volume = ~150 cc), the intrinsic singles rate in the PET energy threshold window will likely be ~15 kHz.

5 Conclusions

In summary, we investigated the basic design and imaging performance of a preclinical PET scanner based on a monolithic annulus of scintillator using Monte Carlo simulations. The

new system is predicted to have high detection sensitivity due to the lack of gaps between detectors. AnnPET's isotropic spatial resolution is ~1 mm and is relatively constant as a function of radial position. Use of monolithic scintillator facilitates estimation of event interaction depth by measuring the shape of the scintillation light spread, which is not possible with detectors that use discrete scintillator elements. Our analysis also indicated that internal scattering of the optical photons can produce minor shifts and distortions of the photon distribution striking the SiPM arrays, resulting in slight errors in calculation of event positioning, and higher errors in DOI measurements. In some cases, the distribution is distorted to the point where it is no longer Gaussian. Though a relatively minor effect, we are investigating new surface treatments to minimize internal photon scattering. The AnnPET detector module described in this investigation could, with minor modifications, be daisy-chained with other modules to create a long scanner that could simultaneously image multiple animals, improving animal throughput. The main limitation of this study is the use of Monte Carlo simulations to predict the performance of the new system. While GATE has been demonstrated to accurately replicate the performance of PET systems undergoing NEMA NU4-2008 evaluation,⁵⁴ the final assessment of AnnPET's performance will require testing on an actual scanner. Therefore, the next step in the project is construction of a prototype system, which is also necessary to assess the hardware requirements for efficient data acquisition and processing.

Disclosures

None of the authors have any competing interests.

Acknowledgments

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